Eesti Maaülikool
Estonian University of Life Sciences

APPLICABILITY OF LOW-COST ELECTROMYOGRAPHS IN ERGONOMIC ASSESSMENT

MADALA MAKSUMUSEGA ELEKTROMÜOGRAAFIDE RAKENDATAVUS ERGONOOMIKALISES HINDAMISES

MÄRT REINVEE

A thesis
for applying for the degree of Doctor of Philosophy
in Engineering Sciences

Väitekiri
filosoofiadoktori kraadi taotlemiseks tehnikateaduse erialal

Tartu 2020

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**Doctoral Theses of the
Estonian University of Life Sciences**

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CONTENTS

LIST OF ORIGINAL PUBLICATIONS

The thesis is based on the following research papers, which are referred to by their Roman numerals:

I **Reinvee, M.**, & Pääsuke, M. (2016). Overview of Contemporary Low-cost sEMG Hardware for Applications in Human Factors and Ergonomics. Proceedings of the Human Factors and Ergonomics Society Annual Meeting, 60(1), 408–412. https://doi.org/10.1177/1541931213601092

II **Reinvee, M.**, Vaas, P., Ereline, J., & Pääsuke, M. (2015). Applicability of Affordable sEMG in Ergonomics Practice. Procedia Manufacturing, 3, 4260–4265. https://doi.org/10.1016/j.promfg.2015.07.412

III **Reinvee, M.**, Aia, S., & Pääsuke, M. (2019). Ergonomic Benefits of an Angle Grinder With Rotatable Main Handle in a Cutting Task. Human Factors: The Journal of the Human Factors and Ergonomics Society, 61(7), 1112–1124. https://doi.org/10.1177/0018720819827184

The contributions of the authors to the papers:

Paper	I	II	III
Study design	**MR**	**MR**, MP	**MR**, SA, MP
Data collection	**MR**	PV, **MR**, JE	SA, **MR**
Data analysis	**MR**	**MR**	**MR**
Manuscript preparation	**MR**, MP	**MR**, PV, JE, MP	**MR**, SA, MP

MR – Märt Reinvee; MP – Mati Pääsuke; PV – Peeter Vaas; JE – Jaan Ereline; SA – Sander Aia.

ABBREVIATIONS

ADC	Analogue-to-digital converter
APDF	Amplitude probability distribution function
CMRR	Common-mode rejection ratio
DIY	Do-it-yourself
EMG	Electromyogram
EVA	Exposure variation analysis
MSD	Musculoskeletal disorder
MVE	Maximal voluntary effort
nEMG	Normalised electromyogram

INTRODUCTION

Musculoskeletal disorders (MSDs) are the main cause of disability and lost productivity in Europe (Bevan, 2015) and the USA (National Research Council, 2001). The economic burden of MSDs (Bhattacharya, 2014; Lee, 1994; Russo, Mariani, Migliorini, Marcellusi, & Mennini, 2015) is perceptible both at the company and national levels, which creates a recognition of the need for prevention. MSDs prevention is the main concern of ergonomics and it requires one to understand the causative factors. In general, there are four major categories of risk factors associated with MSDs (Malchaire, Cock, & Vergracht, 2001): i) personal factors, ii) medical history, iii) psycho-social factors, and iv) occupational risk factors. Out of these only the occupational risk factors are addressable by the means of engineering design. Introducing ergonomic interventions requires an understanding of the specifics of occupational risk factors. Factors like repetitive motions, forceful exertions, non-neutral or awkward postures, and rapid work tempo together with insufficient rest are most often associated with MSD prevalence (Punnett & Wegman, 2004; Putz-Anderson, 1988).

These occupational risk factors are incorporated into the assessment criteria of multiple ergonomic methods and toolkits (David, 2005; Li & Buckle, 1999; Occhipinti & Colombini, 2015; Takala et al., 2010). Surveys about the use of ergonomic methods reveal that ergonomists tend to prefer observational methods to direct measurement methods (Dempsey, Lowe, & Jones, 2019; Dempsey, McGorry, & Maynard, 2005). However, the observational methods are not precise, which may lead to erroneous MSD risk assessment (Diego-Mas, Alcaide-Marzal, & Poveda-Bautista, 2017). Moreover, observations are often carried out only once and the timeframe may not be representative of the whole process, which may further discredit the risk assessment (Punnett & Wegman, 2004). Therefore, multiple authors stress the importance of using quantitative data in risk assessment (Bao, Silverstein, & Spielholz, 2000; Cerqueira, Ferreira da Silva, & Santos, 2019; Dempsey et al., 2000; Ranavolo, Draicchio, Varrecchia, Silvetti, & Iavicoli, 2018), where data is collected with electromyography, force sensing resistors, electrogoniometers, or inertial movements units. In addition, some authors point out that the process of miniaturisation of the sensors in the previously mentioned direct measurement methods (Ranavolo, Draicchio, et al., 2018) makes these sensors also an attractive component for industry 4.0, either as a part of advanced human-machine interaction or as part of a smart occupational risk assessment protocol, which by ensuring healthy workers also enhances productivity (Cerqueira et al., 2019). However, today such opportunities are applied rather in measuring sport performance and in clinical settings than in MSD risk assessment, where the potential is underexploited and applications do not necessarily meet the expectations (Ranavolo, Draicchio, et al., 2018).

One of the unmatched expectations is the cost of such equipment. Respondents in a survey about factors hindering the adoption of wearable sensors at work estimated that companies are willing to pay less than $100 per person for a wearable device (Schall, Sesek, & Cavuoto, 2018). This is not unexpected, as ergonomists tend to prefer inexpensive and effective tools and methods (Dempsey et al., 2019) and it highlights the need to explore low-cost measurement options for MSD risk assessment.

A potential solution is to explore the approach of the maker or do-it-yourself (DIY) movement (Peppler & Bender, 2013) which seems to be the driving force in the development of tools for physical and physiological computing (da Silva, Fred, & Martins, 2014). While the availability of low-cost single-board microcontrollers and web based knowledge have made DIY solutions attainable (Mohomed & Dutta, 2015), the field of physiological computing is fragmented between performance-enhancing exoskeletons, health sensing and telemedicine, and human-computer interaction (da Silva et al., 2014). The fact that the sensors which may be used for quantitative MSD risk assessment are available for the DIY movement at a relatively low price once again highlights the observation of Ranavolo, Draicchio, et al (2018) that these sensors are an unused resource. Therefore, this thesis focuses on one of the direct measurement methods – electromyography – and explores the question whether and to what extent can DIY electromyography be applied in the field of ergonomics.

REVIEW OF LITERATURE

Intrinsic properties and concepts of electromyography

It is common knowledge in the field of ergonomics that electromyography is an area of research specializing in the use of electronic devices to measure and analyse the electrical activity of skeletal muscles. A muscle acts as asynchronous groups of muscle fibres, each group containing <10 to 1000 muscle fibres (Feinstein, Lindegård, Nyman, & Wohlfart, 1955; Loeb & Gans, 1986). A group of muscle fibres together with a controlling α motor neuron form the basic unit of muscle control – a motor unit. The muscle fibres in a motor unit receive commands via action potentials, which essentially are series of rapid depolarisations (from -60…-90 mV to +20…+50 mV) and repolarisations (Luttmann, 1996). The action potentials propagate along the muscle fibres and through conductive body tissues with a speed of 3 to 6 ms^{-1} (Hof, 1984) and can therefore be detected in some distance from the source, either with electrodes inserted into the muscle (intramuscular electromyography) or electrodes attached to the skin above the muscle (surface electromyography). Surface electromyography is preferred in ergonomics due to its non-invasive nature (Marras, 1990). Furthermore, the commercially available DIY electromyography solutions are based exclusively on surface electromyography.

As the action potentials are detected from the surface of the skin, the recorded waveform, i.e. electromyogram (EMG), contains action potentials from multiple motor units. The reason being the fact that electrodes with a diameter of 10 mm are attached to the skin 20 mm apart (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000), while the average motor unit territory is 5 to 10 mm (Buchthal & Schmalbruch, 1980). The detected motor units activate (recruit) in an orderly manner from smallest to largest (Henneman, Somjen, & Carpenter, 1965) and increasing muscular effort is reflected by an increase in the EMG amplitude (Lippold, 1952). EMG amplitude (peak-to-peak) in a resting muscle is in the range of 2 to 10 μV_{pp} (Cram & Kasman, 2010). Meanwhile, in the case of maximum voluntary effort (MVE), the EMG amplitude may reach 4-5 mV_{pp} (Gabriel & Kamen, 2010; Winter, 2009). While the relationship between the EMG amplitude and muscular effort is often reported to be linear, it is not necessarily so because the recruitment of additional motor units is not the only mechanism that regulates the force output. At higher (>50%MVE) levels of muscular effort the frequency with which the muscle fibres are activated (rate coding) becomes increasingly important (Milner-Brown, Stein, & Yemm, 1973). This highlights the importance of spectral characteristics of the recorded EMG. The majority of the EMG spectrum is in the range of 5 to 500 Hz; however, lowering the upper band to 350 Hz is deemed to be acceptable (Merletti & Torino, 2014). Both the frequency and amplitude characteristics of EMG are important factors when choosing equipment for EMG acquisition.

Requirements for the electromyography apparatus

In a robust case one could obtain EMG with an oscilloscope and two metal objects connected to its input (Stegeman & Hermens, 2007), as the electromyograph by its nature is just a very sensitive voltmeter (Cram & Kasman, 2010). For practical reasons, such as the requirements of physiological computing and the specific characteristics of motor unit action potentials, a more sophisticated setup is needed (figure 1).

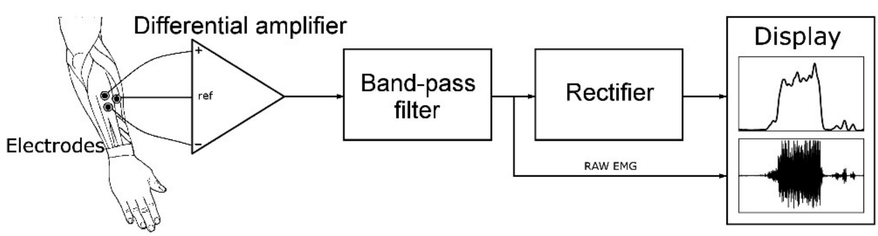

Figure 1. Block diagram of a surface electromyography instrument with two options for output (modified from Cram & Kasman, 2010).

It is evident from figure 1 that the first element in the electromyography instrument (i.e. electromyograph) is a set of electrodes. The electrodes may be the primary factor affecting the signal-to-noise ratio (Gerdle, Karlsson, Day, & Djupsjöbacka, 1999), and it is the skin-electrode interface where the movement artifacts (Merletti, Botter, & Barone, 2016) and thermal noise, V_{th}, (equation 1, Nyquist, 1928) originate:

$$V_{th} = \sqrt{4RkT}, \tag{1}$$

where the k is the Bolzman constant, T is the temperature in Kelvin, and R is the resistance in ohms. The skin-electrode impedance of dry, untreated skin may reach up to 10 MΩ (Cram & Kasman, 2010); however, on average the skin-electrode impedance of 0.8 MΩ with corresponding thermal noise level of 5 μV_{RMS} (root-mean-square) are reported in the literature (Piervirgili, Petracca, & Merletti, 2014). It is evident from equation 1 that the noise level can be reduced by reducing the skin-electrode impedance. This can be achieved with skin treatment. Common treatments include shaving, cleaning with alcohol, and even abrasion (Hermens et al., 2000; Marras, 1990). The skin treatment and other electrode related issues, like their location on the skin (Mesin, Merletti, & Rainoldi, 2009; Nishihara et al., 2010; Takala & Toivonen, 2013; Zipp, 1982), electrode dimensions (Dimitrova, Dimitrov, & Chihman, 1999; Jonas, 1999), inter-electrode distance (Fuglevand, Winter, Patla, & Stashuk, 1992), and the type of electrode gel (Huigen, Peper, & Grimbergen, 2002) are extensively discussed in the literature. However, the same type of electrodes (i.e. standard pre-gelled Ag/AgCl electrodes) are used in the case of both commercial and DIY electromyographs, thus an in-depth analysis of electrode-related matters is outside the scope of this thesis. In contrast, the importance of skin-electrode impedance is discussed further, as it also affects the effectiveness of the differential amplifier.

Most important parameters of the differential amplifier are (Winter, 2009): gain factor, dynamic range, common-mode rejection ratio, and input impedance. There is a general agreement that the amplifier's input impedance should exceed skin-electrode impedance by two orders of magnitude (Bischoff, Fuglsang-Fredriksen, Vendelbo, & Summer, 1999; Cram & Kasman, 2010; Merletti & Hermens, 2004). This means that the amplifier's input impedance should be at least 100 MΩ, although available guidelines do not always specify a value (table 1). This is not surprising, as the ratio of the two impedances is more important than a specific value. The impedance of skin-electrode interface acts as a load that induces a voltage drop in the signal; meanwhile, the amplifier's input impedance will determine the extent of the voltage drop due to the voltage divider effect. Considering that the amplifier's input impedance value is finite, it is clear that the impedance of the skin-electrode interface, which may be reduced with treatment, is the cause of EMG amplitude attenuation. High impedance of the skin-electrode interface may also distort the waveform and induce 50/60 Hz interference in the EMG (Clancy, Morin, & Merletti, 2002).

The 50/60 Hz or powerline interference is removed from the EMG by the amplifier's common-mode rejection ratio (CMRR). As the electrodes are set apart, the propagating motor unit action potentials reach the electrodes at different times. The difference in the signals at the two electrodes is amplified by a factor determined by the amplifier's differential gain (A_d). Meanwhile, the powerline interference, which originates outside the body, reaches the two electrodes at the same time and is amplified by the factor of common-mode gain (A_{cm}). The CMRR or A_{CMRR}, in dB, is defined as the ratio of these two gains and calculated as follows:

$$A_{CMRR} = 20\log\left(\frac{A_d}{|A_{cm}|}\right). \tag{2}$$

The CMRR of an electromyograph is typically in the range of 80 to 120 dB (Örtengren, 1996). It must be noted that the imbalance between the impedance of the two skin-electrode interfaces may render the CMRR useless. Differences of up to 20% (Cram & Kasman, 2010) may be tolerated; however, the imbalance in impedance may result in the amplifier treating the powerline interference as a differential signal. One way to attenuate the powerline interference is to introduce a band-stop filter with a very narrow stopband (i.e. a notch filter). However, this is found to be unsuitable for electromyography (Bischoff et al., 1999; Gerdle et al., 1999), as a significant part of the signal is in the range of 20 to 100 Hz (Gerdle et al., 1999).

Table 1. Summary of amplifier requirements for surface electromyography.

Issue	(Stegeman & Hermens, 2007)	(Merletti & Torino, 2014)	(De Luca, 1997)
Input impedance	'high'	n/a	>100 MΩ
Low-pass filters, Hz	500-1000	>350	500
High-pass filters, Hz	10…20	<10	20
CMRR, dB	n/a	n/a	>80

CMRR – common-mode rejection ratio; n/a – not addressed.

The filters are still used to narrow down the EMG frequency spectrum (table 1). Special attention is paid to the low frequency content inherent to signal artifacts (Örtengren, 1996). These artifacts relate to the movement of electrodes (variation in, or temporary loss of, contact in the skin-electrode interface) or movements of cables (variation of parasitic capacitance in charged loose wires) (Merletti et al., 2016). Another interference located in the EMG low frequency content originates from the heart. The electrocardiogram (ECG) may be detected in the EMG when studying muscles close to the torso. The optimal way to remove the ECG interference and movement artifacts from the EMG is to use a high-pass filter with a corner frequency in the range of 20 to 30 Hz (De Luca, Gilmore, Kuznetsov, & Roy, 2010; Drake & Callaghan, 2006; Redfern, Hughes, & Chaffin, 1993). On the opposite end of the EMG frequency range, low-pass filters are applied in order to avoid aliasing, i.e. sampling related ambiguity. These low-pass filters should have a roll-off of ≥6 dB per octave (Kamen & Caldwell, 1996) and the corner frequency set at half of the sampling rate or slightly below it (Clancy et al., 2002). The corner frequency criteria is determined by the concept of Nyquist frequency. Only 5% of the power in the surface EMG spectrum is located at frequencies >350 Hz (Solomonow, 2000), thus the corner frequency is set at 350 to 500 Hz and the EMG is usually acquired at a sampling rate of 1000 Hz. Oversampling, i.e. using sampling rates significantly higher than required by the Nyquist frequency, is recommended, although the effect of oversampling may be marginal (Durkin & Callaghan, 2005).

Besides the sampling rate, another data acquisition related error is the quantization error. During quantization, the continuous analogue signal is converted into a discrete digital signal. This introduces an ambiguity due to the approximation relating to the resolution of the analogue-to-digital converter (ADC). The quantization error, V_{QE} in volts, is defined as follows:

$$V_{QE} = \frac{1}{2} \cdot \frac{V_{ref}}{2^N - 1}, \tag{3}$$

where V_{ref} is the measurement range in volts, and N is the ADC resolution in bits. The value of V_{ref} depends on the amplifier's gain and the latter needs to be optimised considering V_{QE} and the range of the EMG amplitude. If the lowest possible EMG amplitude (2 μV_{pp}) corresponds to the step value of the digitalised signal, i.e. is equal to $2V_{QE}$, and the highest EMG amplitude (5 mV_{pp}) is equal to V_{ref}, then the dynamic range of the EMG amplitude is equal to $20 \cdot \log_{10}(5\ mV/2\ \mu V) = 68$ dB. The maximum dynamic range (D_{max}) value of an electromyograph is defined as follows (Pozzo, Farina, & Merletti, 2004):

$$D_{max} \approx 6N - 4.8 . \tag{4}$$

From equation (4) it may be observed that a 12 bit ADC, with a D_{max} equal to 67 dB could convert the analogue signal to digital with minimal distortion when the gain is set accordingly. This can be achieved if the amplifier has adjustable gain. The range of 100 to 10,000 is recommended by Winter (2009). However, a 16 bit ADC (D_{max} = 91 dB) could be utilized without significant constraints. Nevertheless, for a while standard AM/FM tape recorders, with a dynamic range 40 to 48 dB (Gerleman & Cook, 1992; Örtengren, 1996), were used to record EMGs. Also, some older digital equipment had 8 bit ADCs (Trontelj, Jabre, & Mihelin, 2004), which allowed for a dynamic range of 43 dB. In this context the 10 bit ADC (D_{max} = 55 dB) of many DIY targeted single-board microcontrollers is not off-putting. It must be noted that using <12 bit ADCs means that either the resolution or the range must be sacrificed to some extent. An alternative to the DIY targeted single-board microcontrollers would be recording the EMG with an external sound card. However, the input impedance of the sound card, 10 kΩ, needs to be matched to the impedance of the skin-electrode interface (Crisp, 2018), a task which may not be executable by everyone and therefore restricts the potential user base. This illustrates the observation of Supuk, Skelin, & Cic, (2014) that DIY electromyographs could hardly compete with the commercial research-grade electromyographs. The latter may cost thousands of euros and therefore trade-offs exist between the cost and accessibility, and between the cost and signal quality. This also indicates the need to monitor the acquired EMG in real time.

Visual inspection of the acquired data may be observed in real time either on a physical display of the electromyograph or nowadays more commonly on a computer or smartphone screen. Independent to the type of display, the EMG is often shown in the form of raw EMG (figure 1, p.12). Raw EMG is the oldest form for presenting EMG, it is an unprocessed, peak-to-peak oscilloscopic (or bipolar) signal (Cram & Kasman, 2010). By observing the raw EMG, one can easily detect signal artifacts and make sure that the signal is in the measurement range; however, it is not easy to estimate the magnitude of a raw EMG with the naked eye. The magnitude of a raw EMG was used frequently in early electromyography studies, often with semi-quantitative scales (Kumar, 1996). Nowadays using processed signals seems to be more common. There are several ways to process raw EMGs, *x(t)*. Two of the most frequently used ways are the root-mean-squared value (V_{RMS}, equation 5) and the average rectified value (V_{AVR}, equation 6):

$$V_{RMS} = \sqrt{\frac{1}{T}\sum_{t=1}^{T} x(t)^2}\ , \tag{5}$$

$$V_{ARV} = \frac{1}{T}\sum_{t=1}^{T} |x(t)|\ . \tag{6}$$

Other common ways of processing are integration (either cyclic for a fixed time/level or over the entire measurement period) and linear envelope (Winter, 2009). In the case of linear envelope, the raw EMG is full-wave rectified and then low-pass filtered. The time constant of a low-pass filter usually ranges between 50 to 300 ms (Solomonow, 2000), which smooths and delays the output. The output follows the muscle's tension, provided that correct filter type and corner-frequency are chosen (Winter, 1996). The filter's corner-frequency (f_c) is calculated as follows (Winter, 1996):

$$f_c = \frac{1}{2\pi\tau}\ , \tag{7}$$

where τ is the muscle's twitch time in seconds. This filter (and also the ones that are used to narrow down the raw EMG spectrum) may be applied either physically in hardware or mathematically in software post-recording. Robertson & Dowling (2003) have provided an equation to calculate filter coefficients and Robbins (2014) has prepared a tutorial to apply the filters mathematically in Matlab and Microsoft Excel. Microsoft Excel is a universal data processing tool, commonly available among standard office software. Therefore, more important than the availability of specific EMG processing software are the knowledge and skills to process and interpret the EMG.

Interpretation of the electromyogram

It is quite obvious that the ability to interpret surface EMG is crucial in the profession of ergonomics, as the design of jobs is often based on the results of electromyographic studies (Ankrum, 2000). In principle, the EMGs that form the results can be obtained by anyone having access to an electromyograph and a few minutes of instruction (Cavanagh, 1974) or using the words of De Luca (1997), 'To its detriment, electromyography is too easy to use and consequently too easy to abuse.' It may be deduced from the previous subchapter that some of the misuses are related to the configuration of the apparatus, while others may be related to the procedures used in attaching the electrodes. In addition to these, there are also issues related to data processing.

EMG is rarely used as its absolute values, i.e. in micro- or millivolts. Rather an EMG is scaled between two comparable anchors, a procedure termed normalisation. Normalisation allows comparison between activities, muscles and subjects (Kumar, 1996; Lehman & McGill, 1999; Marras, 1990) as the absolute EMG values depend on several technical, anatomical, physiological, and procedural factors (Cram & Kasman, 2010; De Luca, 1997). These factors may vary across apparatus, subjects, and muscles (Kumar, 1996). As the effect of these variations is expected to remain constant during a single measurement session, it can be eliminated with the standard min-max normalisation:

$$x_n = 100\left(\frac{x_i - x_{min}}{x_{ref} - x_{min}}\right), \qquad (8)$$

where x_n is the normalised electromyogram (nEMG) in percentages, x_i is the muscle's myoelectric activity during a task in volts, x_{min} is the muscle's myoelectric activity during rest in volts, and x_{ref} is the muscle's myoelectric activity during the reference contraction in volts. The reference contraction is often the subject's maximal voluntary effort (MVE) and nEMG is therefore expressed in %MVE. However, there are also reasons to prefer a submaximal effort to maximal: i) obtaining a true MVE depends on the subject's motivation (Kumar, 1996), and ii) the nonlinearity related to the increased firing frequency at higher percentages of the MVE (Milner-Brown et al., 1973). In the case of submaximal reference, the effort is either some percentage of the MVE or a fixed load, e.g. holding a load of 5 kg. The choice of how and whether to normalise depends on the application and the reason why electromyography is used.

There are four main reasons to use electromyography in ergonomics, as an EMG provides insight about (Marras, 1990): i) the duration of the muscular load, ii) the relative level of the muscular load, iii) muscular fatigue, and iv) muscular force, although with limitations. Some of these rely on EMG amplitude and require normalisation while others are based on the changes in EMG frequency spectrum.

The simplest use of electromyography is to evaluate the duration of muscular load. Durations of muscular load and rest periods play a major role in ergonomic investigations, as the effect of the work-rest pattern on the endurance limit in long-term intermittent contractions has been known for a long time (Björkstén & Jonsson, 1977). The rest periods in the work-rest pattern are also visible in EMGs. While longer rest periods in EMGs may be observed with the naked eye, the literature seems to focus on very short rest periods named EMG gaps and defined as periods of ≥0.2 s with myoelectric activity below 0.5%MVE (Veiersted, Westgaard, & Andersen, 1990). EMG gap frequency tends to decrease in the case of higher mental workload (Schleifer et al., 2008) and infrequent occurrence of EMG gaps is associated with MSDs (Hägg & Åström, 1997).

The risk of MSD occurrence is also associated with muscular fatigue (National Research Council, 2001). It is possible to evaluate muscular fatigue with electromyography. However, instead of the direct measures of mechanical fatigue one can use indicators that reflect the change in the electrical manifestations of fatigue (Hägg, Luttmann, & Jäger, 2000). Some of these indicators are amplitude based (e.g. V_{RMS}, V_{ARV}) while others are based on the frequency spectrum (e.g. the mean and median frequency). The mean frequency (f_{MNF}, equation 9) is in the range of 70 to 130 Hz and the median frequency (f_{MDF}, equation 10) is in the range of 50 to 110 Hz (Merletti, Rainoldi, & Farina, 2001).

$$f_{MNF} = \frac{\int_0^{\frac{f_s}{2}} f P(f) df}{\int_0^{\frac{f_s}{2}} P(f) df} . \tag{9}$$

$$f_{MDF} = \frac{1}{2} \int_0^{\frac{f_s}{2}} P(f) df \cdot \tag{10}$$

Obtaining correct f_{MNF} and f_{MDF} values depends on the sampling frequency *(f_s)* as the power spectral density, i.e. *P(f)*, is based on the fast Fourier transform. Thus obtaining EMG when it is band-pass filtered to the range of 10 to 350 Hz requires $f_s \geq 700$ Hz, meanwhile the linear envelope of EMG amplitude may be obtained with sampling rates between 50 to 100 Hz (Merletti & Torino, 2014). There are also differences in the interpretation of amplitude and frequency based indicators. While f_{MNF} and f_{MDF} will decrease with progressing fatigue, the opposite is true in the case of V_{RMS} and V_{AVR} (Cifrek, Medved, Tonković, & Ostojić, 2009; Merletti, Lo Conte, & Orizio, 1991). Regardless of the indicator, it is advised to test the fatigue state with contractions lighter than the actual task, e.g. 5%MVE (Søgaard, Blangsted, Jørgensen, Madeleine, & Sjøgaard, 2003). The reason to choose such a low effort for test contractions is two-fold. Firstly, as the motor units are recruited in an orderly manner from smallest to largest (Henneman et al., 1965), a low effort test contraction ensures that the motor units that are active in the test contraction were fatigued during the work task. Secondly, higher efforts (>15%MVE) may induce fatigue themselves (Rohmert, 1973).

The 15%MVE is often used as design criteria, clearly indicating the importance of relative effort. In ergonomics this relative effort may need to be monitored from several hours to the length of a work day (Hof, 1984). It may be observed from the left of figure 2 that fast interpretation on long-term recordings is not a simple task and data reduction is needed.

There are two methods used to allow grasping the data at a glance (Hägg et al., 2000): amplitude probability distribution curve (APDF) and exposure variation analysis (EVA). APDF (figure 2, right) describes the profile of the relative muscular load during a work period and the following criteria are used to evaluate the curve (Jonsson, 1982): the 10^{th} percentile must not exceed 5%MVE, the 50^{th} percentile must not exceed 14%MVE and the 90th percentile must not exceed 70%MVE. In contrast, EVA (Mathisassen & Winkel, 1991) does not have clear assessment criteria and it may be difficult to use the results in risk assessment (Westgaard, 2000). The difficulties arise from the fact that the EVA result is a data matrix, as EVA adds a third axis to APDF (the distribution of periods in which the amplitude stays on certain levels). Hägg & Åström (1997) have shown that this third axis allows using EVA to analyse EMG gaps. All in all, EVA is not a method to be used by non-researchers, as interpreting and visualising the results is challenging (Anton, Cook, Rosecrance, & Merlino, 2003).

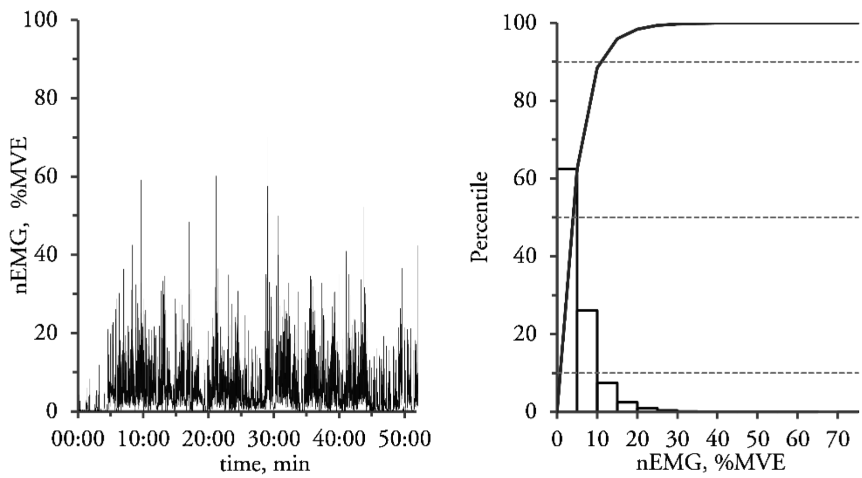

Figure 2. Time series and amplitude probability distribution function; a 50-minute time series of a normalized electromyogram (nEMG) on the left; and corresponding amplitude distribution histogram and amplitude probability distribution curve on the right (modified from Reinvee & Mrugalska, 2019).

It is also challenging to use EMG in force prediction. As mentioned previously, an increasing effort is reflected in EMG by changes in amplitude due to the recruitment of additional motor units (Henneman et al., 1965) and also by changes in EMG frequency characteristics (Milner-Brown et al., 1973), neither being necessarily linearly related to force. This issue has been addressed in several ways; some authors have used multiple nonlinear regressions (Duque, Masset, & Malchaire, 1995) and some have used non-linear normalisation (McDonald, Sanei, & Keir, 2013; Potvin, Norman, & McGill, 1996) while others have carried out series of calibrations (Hoozemans & van Dieën, 2005; Keir & Mogk, 2005). Hof (1984) points out that the complicated EMG-force relationship originates from the complexities of muscle mechanics.

In the case of a single posture, an EMG from a single channel can adequately reflect the EMG-force relationship by a linear (Duque et al., 1995) or curve-linear relationship (Gurram, Rakheja, & Gouw, 1995). When the EMG-force relationship in multiple postures is needed, then it is necessary to use models with three or more muscles (Hoozemans & van Dieën, 2005; Keir & Mogk, 2005).

Whatever the case, reaching valid conclusions requires a fundamental understanding of electrophysiology and anatomy (Cavanagh, 1974; Clarys, 2000). Additional skills and knowledge may be necessary when applying DIY electromyographs into practice (Reinvee & Mrugalska, 2019).

Applications of electromyography

In practice, surface electromyography is used in measurement, feedback, and control applications. It may be already gathered from the literature analysis above that in ergonomics EMG is used in risk assessment (Granzow et al., 2018; Merino, da Silva, Mattos, Guimarães, & Merino, 2019; Ranavolo, Varrecchia, et al., 2018), job design and organisation (Gaudez, Wild, & Aublet-Cuvelier, 2015; Maciukiewicz, Cudlip, Chopp-Hurley, & Dickerson, 2016), hand tool and instrument design (Agostinucci & McLinden, 2016; Conner & Irwin, 2009; Douphrate et al., 2017; Duke, Mirka, & Sommerich, 2004; Steinhilber et al., 2017), and fatigue assessment (Fattorini et al., 2017; Jansen et al., 2012; Kimura, Sato, Ochi, Hosoya, & Sadoyama, 2007; Lin, Liang, Lin, & Hwang, 2004). There are several reasons to use low-cost DIY electromyographs for the objectives listed above. Firstly, the high initial capital investment keeps the commercial electromyographs out of reach for many potential users and even laboratories (Dempsey et al., 2019; Supuk et al., 2014). Secondly, there is a risk that the high-cost apparatus becomes obsolete or is infrequently used (Stojanovic, Hagara, Ondracek, & Caplanova, 2015). Thirdly, there is a risk of equipment damage when collecting data outside the laboratory (Walters, Kaschinske, Strath, Swartz, & Keenan, 2013). Fourthly, commercial devices often function as 'black-boxes', without the opportunity to process data independently, and thus pose constraint for research (Baeyens et al., 2018). Fifthly, an EMG in isolation has limited value (Cavanagh, 1974); another relatively high capital investment may be required in order to synchronise the EMG with other types of sensors.

Synchronising the EMG with other types of sensors or rather with actuators is mandatory in the case of biofeedback applications. The applications fall into three main categories (Cram, 2005): down-training (i.e. systemic relaxation), up-training, and coordination training. All three are important strategies when training workers to use their muscles more efficiently during computer use (Gaffney, Maluf, & Davidson, 2016; Madeleine, Vedsted, Blangsted, Sjøgaard, & Søgaard, 2006; Peper et al., 2003; Vedsted, Søgaard, Blangsted, Madeleine, & Sjøgaard, 2011), assembly (Faucett, Garry, Nadler, & Ettare, 2002; Parenmark, Engvall, & Malmkvist, 1988), and manual material handling (Agruss, Williams, & Fathallah, 2004) in order to prevent MSDs. These applications do not propose high requirements for EMG acquisition as feedback is often triggered when a threshold is exceeded. The requirements may be somewhat higher in the case of control applications.

From the ergonomics perspective, there is a significant interest in wearable exoskeletons which would allow to reduce the risk of MSDs (de Looze, Bosch, Krause, Stadler, & O'Sullivan, 2016). An exoskeleton might be successfully controlled with the input form electromyography (Hara & Sankai, 2010) and it has been shown to improve workers' performance and to reduce perceived fatigue (Tan et al., 2019). However, recent reviews (Fox, Aranko, Heilala, & Vahala, 2019; Theurel & Desbrosses, 2019) exhort to maintain scepticism, as the effects of exoskeleton induced changes in kinematics are not fully known. This is only natural, as exoskeletons are still an emerging technology and solutions for industrial purposes are under development (de Looze et al., 2016). There are several examples where low-cost electromyography is used to improve and develop strategies for electromyography-controlled exoskeletons or prostheses (Hassan, Abou-Loukh, & Ibraheem, in press; Lu et al., 2019; Russo, Fernández, & Rivera, 2018; Secciani, Bianchi, Meli, Volpe, & Ridolfi, 2019). Using a DIY electromyography-based exoskeleton in the workplace would be an unthinkable and irresponsible practice; however, the fact that low-cost electromyography has been successfully adopted in the research of exoskeletons and prostheses encourages testing their potential to be used in ergonomic measurement of feedback applications.

AIM OF THE STUDY

The available literature acknowledges at least partially the linear relationship between force exertion and EMG amplitude. Thus the indirect assessment of relative effort or force exertion via electromyographic measurement is appreciated in the ergonomic assessment. Although electromyography is desired for multiple reasons in ergonomic assessment, so far the availability of electromyography has been restricted by the cost of the apparatus. This is not surprising as the commercial electromyographs may cost several tens of thousands of euros (Supuk et al., 2014). Recently the cost restriction has been addressed by the trend to increase the availability of tools for physiological computing. These tools can be used to construct apparatus with electromyography functionality. The cost of such apparatus is usually just a fraction of what a commercial device would cost, often not exceeding one or two hundred euros. This low-cost apparatus is mainly used in the development of limb prostheses; meanwhile the potential to use the apparatus in ergonomic assessment is unknown.

The aim of the thesis was to assess the applicability of contemporary low-cost electromyographs for using in ergonomic assessment.

In order to achieve the aim, the following tasks were set:

1. Analyse the technical characteristics of contemporary low-cost electromyographs and their compliance with the hardware requirements of electromyography in ergonomics (expanded from Paper **I**).
2. Evaluate the quality of contemporary low-cost electromyographs in the laboratory and in the field (Papers **I-III**)
3. Propose, create and test applications of contemporary low-cost electromyographs for ergonomic assessment of hand tools (Papers **I-III**).

MATERIALS AND METHODS

Apparatus

Electromyography. EMGs were obtained either with a low-cost DIY electromyograph (**I-III**) or a commercially available research grade electromyograph ME6000 (Mega Electronics, Finland) (**II**). The technical characteristics of ME6000 are as follows: measuring range ±8192 μV, resolution 1 μV, gain 1, CMRR 110 dB, and band-pass filtering 8…500 Hz. The low-cost DIY electromyograph was based on an EMG compatible bioamplifier (table 2) that was connected to either an Arduino Leonardo (Arduino LLC, Italy) or BITalino (PLUX wireless biosignals S.A., Portugal) microcontroller. The measurement range of the Arduino Leonardo is 5 V and the resolution of the A/D converter 10 bit. BITalino's measurement range is 3.3 V and the resolution of the A/D converter is 10 bit. In **II** & **III** the sampling rate was set to 1000 Hz, in **I** the sampling rate was 100 Hz.

Table 2. Characteristics of EMG compatible bioamplifiers (data from I).

Amplifier	Signal	Filters, Hz	Z_{in}	CMRR, dB	Gain	Paper
BITalino v.151015	raw	BP 10-400	100 GΩ	110	1000	I, III
FlexVolt Shield	raw	none	n/a	100	2336	I
Olimex Shield EKG/ EMG	raw	LP f_c = 40	n/a	80	205..3595	I, II
MyoWare	raw & linear envelope	none/ LP f_c = 2	110 GΩ	110	0.002..20100	I
Muscle Sensor v3	linear envelope	LP f_c = 2	n/a	90	0.002..20700	II

Z_{in} – input impedance; CMRR – Common-mode rejection ratio; BP – band-bass; LP- low-pass; f_c – cut-off frequency; n/a – no value available.

In all experiments bipolar dual Ag/AgCl disposable electrodes with an inter-electrode distance of 2.0 cm (Noraxon Inc, Scottsdale, USA) were used. Before attaching the electrodes to the skin, an appropriate area on the skin was cleaned with alcohol and shaved when necessary. The area was identified based on the location in an anatomy atlas (Cram, Kasman,

& Holtz, 2011), the exact locations to attach the electrodes were located by palpation and confirmed visually by inspecting the data of muscle contractions.

Dynamometry. In most cases (**I**, **III**) grip force data was collected with an electronic I-type dynamometer (Vernier Software & Technology, USA), circumference 155 mm, which was connected to an Arduino Uno microcontroller (Arduino LLC, Italy). In the second laboratory study (**II**) an I-type electronic dynamometer (Neurosoft, Russia) with a circumference of 135 mm was used. The dynamometer was directly connected to a personal computer and data was collected with the software provided by the manufacturer of the dynamometer.

Feedback system. Either visual (**I**, **III**) or a combination of visual and auditory (**II**) feedback was used to assist the subjects in following the test procedure. In the case of only visual feedback, three coloured LEDs (red, orange, and green) were connected to an Arduino Nano microcontroller (Arduino LLC, Italy). The LEDs were turned on and off in a pre-programmed sequence. The investigator launched the sequence with a remote which controlled a 315 Mhz T4 receiver (Adafruit, USA) connected to the microcontroller. In the case of the combination of visual and auditory feedback, a metronome was used to indicate the pace, which allowed the subject to follow the Caldwell regimen (Caldwell et al., 1974). Meanwhile the exertion level (25, 50 and 75%MVC) was displayed on the computer screen via the НС-Психотест software (Neurosoft, Russia).

Subjects

The subjects in the laboratory studies (**I**, **II**) were recruited from the population of local universities. In contrast, the subjects in the field study (**III**) were recruited from the population of local metalwork companies and self-employed maintenance workers. All subjects were males and the characteristics of the subjects are shown in table 3.

Table 3. Characteristics of the subjects, mean ± SE.

Paper	*n*	Age, yrs	Height, cm	Body mass, kg	Grip force, N
I	6	25.8 ± 2.5	184.5 ± 5.1	75.0 ± 1.3	466 ± 36
II	10	24.9 ± 0.7	180.8 ± 1.1	80.4 ± 3.3	397 ± 13
III	11	30.5 ± 2.1	181.0 ± 2.2	84.1 ± 2.8	421 ± 73

The subjects' personal data management conformed to the tenets of the Declaration of Helsinki.

Experimental design

The applicability of the low-cost electromyograph was analysed in three stages. **In the first stage** internet search was conducted to identify the available products for EMG acquisition and the technical properties of such products were compared to the functional requirements described in the literature. The following criteria were used in the internet search: 1) cost below $150; 2) the product is freely available in an online store; 3) absence of negative feedback in the comments section of the online store and 4) the product does not require a significant amount of tinkering or DIY approach. The identified products classify as bioamplifiers and their available technical properties are summed in table 2 (p. 25).

Low-cost single-board microcontrollers were then studied in order to determine their capability to sample EMGs. In particular, the amount of channels that can be used for linear envelope ($f_s < 100$ Hz) and raw EMG ($f_s > 700$ Hz) acquisition were tested. In order to represent results that are obtainable by a wide user base, low level programming was avoided and a simple timer by Mellis, Stoffregen, Fitzgerald, & Guadalupi (2005) was used. Sampling rates of 100, 800 and 1000 Hz were tested, and the acceptance criteria was set to measurement period fluctuations of <10 μs.

In the second stage, the selected bioamplifiers were used in the DIY approach electromyographs and tested in the laboratory (**I**, **II**) or in the field (**III**). To some extent, a common approach was used both in the laboratory and in the field. The common procedural aspects were:

1) data of exertions related to grip strength were collected in accordance to the Caldwell regimen (Caldwell et al., 1974). In short, the subjects were instructed to increase their effort smoothly without jerking movements during one second and then hold their maximum voluntary effort for four seconds. In order to increase compliance with the instructions, the subjects were trained to follow the regimen and either an auditory (**II**) or visual (**I**, **III**) feedback system was used to indicate the pace.
2) the exertions of maximal voluntary effort were used to normalise EMG data. Eq. 8 was used to calculate the normalised EMG value.

3) the subjects were trained to follow the test procedure and to understand the feedback system.

Only one muscle (*m. flexor digitorum superficialis*) was investigated in the laboratory studies (**I**, **II**), as the focus was on the performance of the low-cost apparatus. Based on prior knowledge, the relationship between the forearm muscle's EMG amplitude and grip force exertion was expected to be at least partially linear. Initially (**I**) the linearity was tested in a small (n = 6) study group with freely chosen submaximal efforts. The tests were then repeated (**II**) in a larger (n = 10) study group at clearly indicated effort levels (25%, 50% and 75% of maximum voluntary effort). Also, the performance of the low-cost DIY-electromyographs was compared to a commercially available research grade electromyograph (**II**).

In contrast to the laboratory studies (**I**, **II**), the field study (**III**) focused on the practice of ergonomics. Thus the number of muscles investigated was increased to three (*m. flexor carpi ulnaris, m. flexor carpi radialis,* and *m. extensor digitorum*) and only one low-cost electromyograph, BITalino (PLUX wireless biosignals S.A., Portugal), was used. The field study (**III**) aimed to increase the body of knowledge about the ergonomic assessment of angle grinders. Specifically, an angle grinder with a rotatable main handle was studied. The rotatable main handle allows an angle grinder operator to cut horizontally positioned material in three wrist postures (figure 3).

Figure 3. Postures: A – the bow of the main handle is perpendicular to the abrasive disk, the trigger is activated with four fingers, the posture causes significant wrist flexion; B – same as posture A, but the trigger is activated with the thumb, the posture causes just noticeable wrist extension; C – the bow of the main handle is in line with the abrasive disk, the trigger is activated with four fingers, the wrist is in near-neutral position (reproduced from **III**).

These three postures, A, B and C, were compared in a simulated work sequence by the forearm muscles' myoelectric activity.

Data was obtained from a group (n = 11) of metal- and maintenance workers. The subjects performed a sequence of three simulated work tasks. The three tasks were as follows: 1) holding the angle grinder in a static posture; 2) starting the angle grinder and holding the device while the grinding disk is running at full speed 3) cutting three details from a Ø10 mm steel rod with Rockwell hardness of 77.2 HRB (SE 0.4). The steel rod was firmly secured between the jaws of metalworking vices. The height of the metalworking vices dictated the fixed height of the steel rod, while the length of the cuts from the steel rod was freely chosen by the subjects. Although the height of the steel rod was fixed and the stature of the subjects varied, the interaction of these factors was not analysed. The focus was solely on the effect of postures A-C to the forearm muscle load. In order to distinguish the tasks in the sequence, a single axis accelerometer was connected to the EMG acquisition device. The dataset of the accelerometer was band-pass filtered according to the scheme in figure 4, which allowed to obtain comparable samples from the EMG dataset.

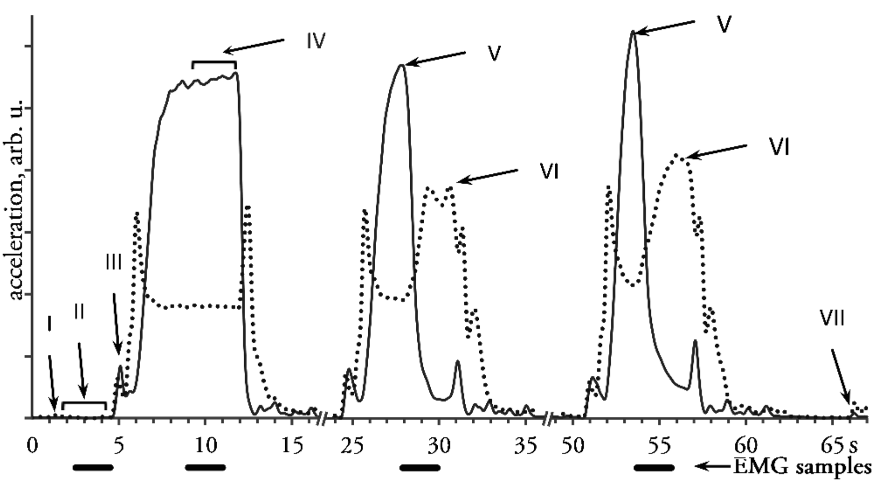

Figure 4. Data processing scheme in the field study: solid line – acceleration band-pass filtered in range 100-110 Hz; dotted line – acceleration data, band-pass filtered in range 60-95 Hz; I – picking up the angle grinder; II – static holding of the device; III – pressing the trigger; IV – full speed, no load; V – start of the cut; VI – releasing the trigger; VII – putting the device away (modified from **III**).

In the third stage the accumulated information and experience from the first and second stage was analysed to discuss the potential applications of low-cost electromyography in ergonomics. An approach described by

Stanton (2006) was used to analyse the hardware and software solution for the applications.

Data processing

The raw EMG data was processed either in Microsoft Excel (**I**, **II**) or with custom Matlab (MathWorks, Inc., USA) scripts (**III**). The Matlab scripts were used to remove the DC-offset, calculate the linear envelope from raw data and to visualise the data. For linear envelope a Butterworth zero-lag, second-order low-pass filter, with corner frequency set to 2 Hz was used to smooth the data. Microsoft Excel worksheets (**I**, **II**) were used for similar purposes as the Matlab scripts (**III**); however, the V_{RMS} (Eq. 5) or V_{ARV} (Eq. 6) values were used instead of linear envelope.

The statistical analysis was conducted in the software R (R Development Core Team). First, the normality and homoscedasticity of the data were tested with the Shapiro-Wilk and the Fligner-Killeen tests. Based on the results either a parametric (ANOVA with Tukey HSD *post hoc* test) or a non-parametric test (Friedman and Wilcoxon Rank-Sum test with Benjamini & Hochberg procedure for pairwise comparisons) was used.

Data is reported as mean ± standard error (SE), the level of statistical significance was set to $p < 0.05$.

RESULTS

Laboratory studies

Firstly, the characteristics of the low-cost single-board microcontrollers that can be interfaced with the bioamplifiers (listed in table 2) were studied. It is apparent that all the microcontrollers can be used to register lineal envelope while only some can be used for raw EMG acquisition (table 4).

Table 4. Characteristics of low-cost single-board microcontrollers; (unpublished data, collected for **I**).

Microcontroller	FS, V	ADC resolution	Available channels at given f_s		
			100 Hz	800 Hz	1000 Hz
Arduino Uno	5.0	10 bit/4.9 mV	6/6	1/6	0/6
Arduino Mega	5.0	10 bit/4.9 mV	16/16	1/16	0/16
Arduino Leonardo	5.0	10 bit/4.9 mV	6/6	2/6	1/6
Arduino Due	3.3	12 bit/0.8 mV	12/12	6/12	4/12
BITalino	3.3	10 bit/3.2 mV	4/4	n/a	4/4

FS – full scale; increasing the sampling rate (f_s) reduces the number of usable analogue channels (i/j), where i is the number of channels available at given f_s and j is the total number channels; n/a - sampling rate not available.

Secondly, the linearity in the relationship between the nEMG and relative grip force was studied using a single EMG compatible bioamplifier connected to the Arduino Leonardo microcontroller board. The result (figure 5) was fairly linear.

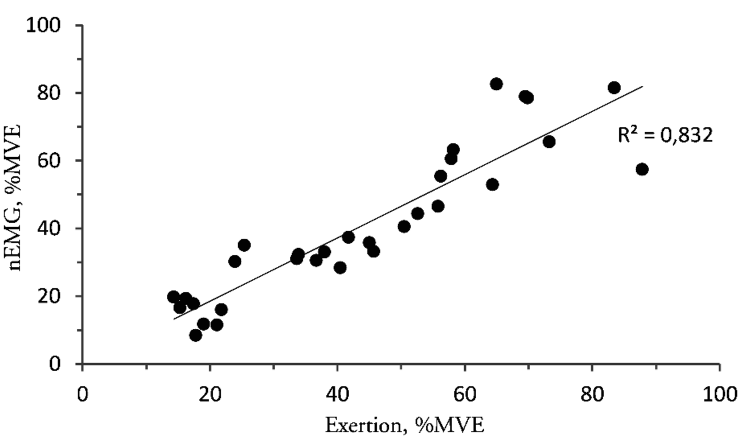

Figure 5. Relation between the normalised electromyogram (nEMG) amplitude of *m. flexor digitorum superficialis* and grip force exertion; 100% is equal to maximal voluntary effort i.e. MVE; (n = 6, freely chosen submaximal efforts, adapted from **I**).

The relation between relative effort and nEMG was further studied in a larger ($n = 10$) study group and at clearly indicated effort levels (25%, 50%, and 75%MVE). Also, the performance of two DIY electromyographs was compared to the performance of a commercially available research grade electromyograph (figure 6).

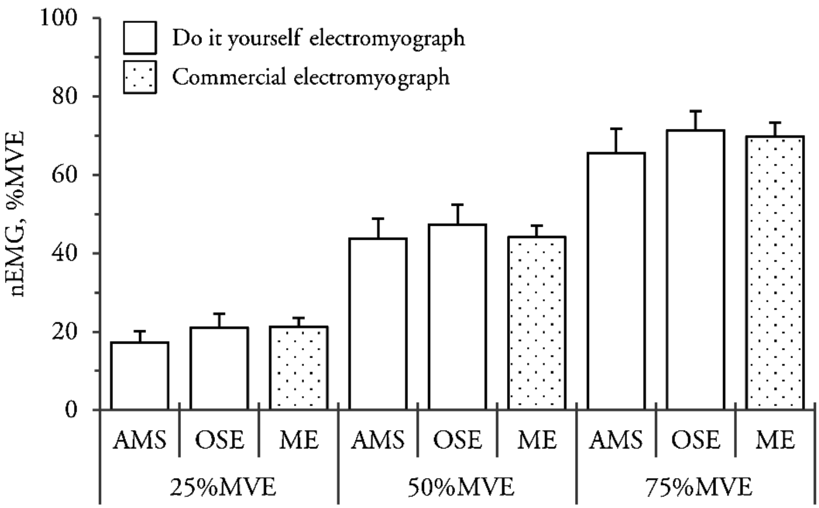

Figure 6. The effect of the apparatus on the normalised electromyogram (nEMG) amplitudes of *m. flexor digitorum superficialis*, on three submaximal effort levels (mean + SE); 100% is equal to maximal voluntary effort i.e. MVE; AMS - Advancer Technologies' Muscle Sensor v3; OSE - Olimex Shield EKG/EMG; ME - electromyograph ME6000 (adapted from **II**).

The effect of the apparatus on the nEMG was not statistically significant at any of the three effort levels. The lowest p value was found at the level of 25%MVE – $F(2, 27) = 0.6$, $p = 0.555$.

Field study

In the field study (**III**), the nEMG amplitudes were compared within the muscle and between postures or tasks (figure 7). Firstly, the effect of posture on the nEMG was analysed during different tasks. In the case of *m. flexor carpi ulnaris* and *m. flexor carpi radialis*, the posture did not have a statistically significant effect on the nEMG in any of the tasks studied. In the case of *m. extensor digitorum*, the nEMG was statistically significantly affected by posture, $\chi^2(2) = 15.000$, $p = 0.001$, while holding the angle grinder.

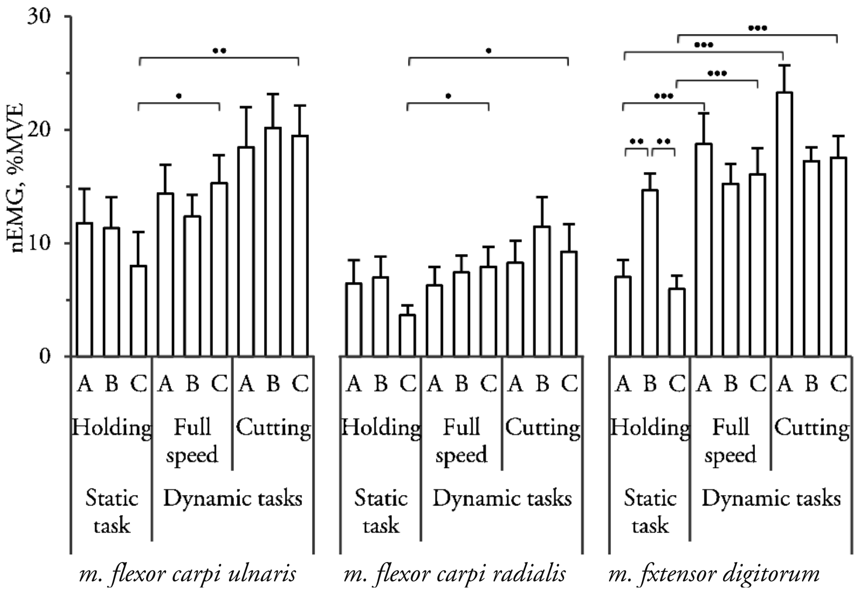

Figure 7. Normalised electromyogram (nEMG) amplitudes according to posture (A-C) and task (mean + SE); 100% is equal to maximal voluntary effort i.e. MVE; see figure 3 for postures A-C ; * - $p < .05$, ** - $p < .01$, *** - $p < .001$ (adapted from **III**).

Secondly, it was analysed if the task had an effect on the nEMG within a posture. In the case of posture A, only the nEMG of *m. extensor digitorum* showed a statistically significantly effect of task, $\chi^2(2) = 15.800, p < 0.001$. In the case of posture B, both *m. flexor carpi ulnaris*, $\chi^2(2) = 7.400$, $p = 0.025$, and *m. flexor carpi radialis*, $\chi^2(2) = 7.800$, $p = 0.020$, showed a statistically significantly effect; however, the Benjamini & Hochberg procedure in pairwise comparison rendered these probabilities to non-significant ($p > 0.079$). Only in posture C a statistically significantly effect of task was noted in all three muscles.

Applications

Data in table 4 and figures 5-7 indicates that low-cost DIY electromyographs may be used for electromyography based biofeedback. Hardware (figure 8) and software (figure 9) structures to construct and program such devices were proposed in paper **I**. The modifications on the figures are partially based on the analysis presented in Štrik-Ott (2019) – a master's thesis supervised in the framework of this PhD thesis.

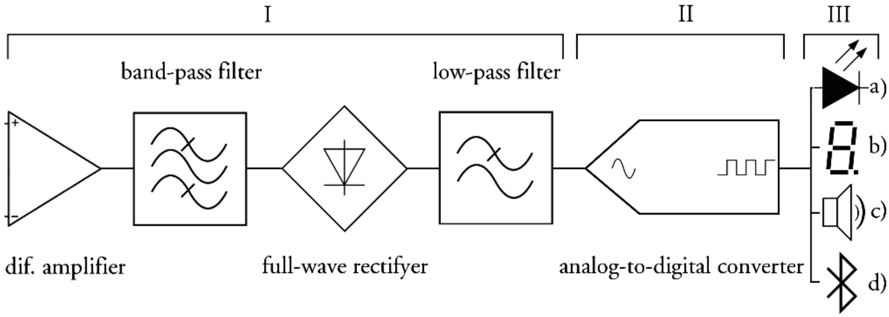

Figure 8. Structure of an electromyography feedback device; I – EMG functionality board, II – single board microcontroller; III – one or multiple actuators: a) LED, b) numeric display c) piezo speaker d) Bluetooth; (modified from **I**).

0. Provide electromyography feedback
Plan 0: Do 1 then 2 then 3 then 4 then 5 then repeat either 2-5 or 2-4 or 2&3&5.
  1. Start device
  Plan 1: Do 1.1 then 1.2 and/or 1.3.
    1.1. Set reference values to default
    1.2. Set audio feedback pitch range
    1.3. Set display range
    1.4. Take 20 readings from electromyography sensor
    1.5. Calculate the average
    1.6. Set 1.5 to rest EMG value
  2. Measure
  Plan 2: Do 2.1 then 2.2 then 2.3.
    2.1. Reset the reading counter
    2.2. Take 20 readings from electromyography sensor
    2.3. Calculate the average
  3. Normalise
  Plan 3: Do 3.1 then 3.2 then 3.3 then 3.4 and/or 3.5.
    3.1. Read potentiometer for reference value
    3.2. Set new reference value
    3.3. Scale 2.3 between 1.6 and 3.2
    3.4. Scale 3.3 to pitch range
    3.5. Scale 3.3 to display range
  4. Play tone
  Plan 4: Do 4.1 then 4.2.
    4.1. Play pitch value of 3.4
    4.2. Wait 10 ms
  5. Display LEDs
  Plan 5: Do 5.1 then 5.2 then 5.3.
    5.1. Set all LEDs off
    5.2. Show no. of LEDs corresponding to the value of 3.5
    5.3. Wait 100 ms

Figure 9. Hierarchical list for the software of an electromyography feedback device; (modified from **I**).

The proposed software structure (figure 9) utilises actuators ‘a’ and/or ‘c’ from the figure 8. The numeric values in figure 9, i.e. the number of readings to take in order to smooth data and the delay to reduce the flickering of the LED display are a result of trial and error and thus subjects to change in case of different hardware settings or studied muscles. A prototype, following the structures in figures 8 and 9, was built and tested (https://youtu.be/57n6Hbv5WEo). The two-channel prototype cost 120 euros to make, where the specific cost of a single EMG channel was 50 euros. Having two channels allows to demonstrate the effect of posture (flexion-extension test) and compare muscle loads (grip force exertion test); both are important for ergonomic assessment of hand tools. The dimensions 90 x 60 x 30 mm make the device portable and allow to hold the device with one hand when necessary.

DISCUSSION

Analysis of compatibility

For any given system the usability of the system may be described by its effectiveness and efficiency (ISO 9241-11:2018). Effectiveness in terms of low-cost electromyography refers to the technical capability to detect, record, and store or display the myoelectric activity. To a large extent the effectiveness may be evaluated by comparing the requirements (table 1, p. 14) and characteristics of EMG compatible bioamplifiers (data collected for paper **I**, presented in table 2, p. 25).

In general, a bioamplifier is required to attenuate signal artifacts and amplify the low-voltage EMG amplitude. As the low-voltage EMG needs to be amplified prior to analogue-to-digital conversion, an amplifier is required to have high input impedance in order to avoid voltage drop by the voltage divider effect. As a rule of thumb, the amplifier's input impedance should be at least 10 to 100 times greater than the skin-electrode impedance (Cram & Kasman, 2010; Merletti & Hermens, 2004). The skin-electrode impedance of untreated skin is expected to be near 1 MΩ (Piervirgili et al., 2014), thus the input impedance should exceed 100 MΩ. It is evident from table 2 that according to the available information, the input impedance of the selected bioamplifiers is at least three orders of magnitude greater than the recommended value of 100 MΩ. Similarly the bioamplifiers' common mode rejection ratio (CMRR) is greater than or equal to 80 dB, which means that the capacity to attenuate external noise in EMG exceeds the recommended value (table 2).

As the selected bioamplifiers satisfy the general criteria for biosignal acquisition, the next step of the evaluation is to compare the properties of the selected bioamplifiers with the specific requirements of the EMG, namely its amplitude and frequency characteristics. EMG amplitude is expected to be in the range 2 to 10 μV_{pp} in the case of resting muscles (Cram & Kasman, 2010) and known to reach 4 to 5 mV_{pp} in the case of maximum voluntary contraction (Gabriel & Kamen, 2010; Merletti & Hermens, 2004; Winter, 2009). In addition, the spectrum of EMG is known to contain the majority of its power in the range of 5 to 500 Hz (Merletti & Torino, 2014). Acquiring an EMG while maintaining its amplitude and frequency characteristics is not an issue that can be successfully addressed just by the bioamplifier's characteristics. Rather it is a matter of fit between the characteristics of the bioamplifier (table 2) and the low-cost single-board microcontroller (table 4).

In the case of the EMG amplitude characteristics, two issues need to be considered, namely range and resolution. Rendering the EMG amplitude without distortion requires a fit between the amplifier's gain and the microcontroller board's measurement range. Theoretically, a full scale measurement rage of 5 V requires a gain below 1000 and a full scale measurement rage of 3.3 V requires a gain rate below 660 (**I**). This implies that the bioamplifiers with fixed gains in table 2 may clip the signal due to overamplification and distort data. However, in practice the EMG amplitude depends on several issues. In addition to the intermuscular differences, the maximum of EMG amplitude also depends on several factors, such as the width of the fatty tissue between the muscle and the skin, posture, and inter-electrode distance (Cram & Kasman, 2010; De Luca, 1997). Although with caution, this suggests that that there may be muscles and procedures which allow recording EMGs of maximum voluntary contractions with the above-mentioned fixed gain bioamplifiers. A bit more flexible alternative to the gain problem is an amplifier with adjustable gain. Winter (2009), suggests adjustability in the range of 100 to 10,000. Two out of three adjustable gain bioamplifiers in table 2 allow gain adjustment in this range. Therefore, contemporary low-cost electromyography allows addressing the range issue fully in the case of configurations with adjustable gain bioamplifiers and conditionally in the case of configurations with fixed gain bioamplifiers. The issue of resolution is more complex than the gain issue. In addition to the amplifier's gain and the microcontroller board's measurement range, the ADC's resolution must also be considered. The fit of these three factors will determine: i) whether EMG amplitude of the resting muscle is distinguishable from noise and ii) how small changes in activity can be interpreted. The contemporary low-cost microcontroller boards usually have either 10 or 12 bit ADC resolution, which is clearly inferior to the ≥16bit ADC resolution of contemporary commercial electromyographs. However, Trontelj et al. (2004) point out that in the past ADCs with 8 bit resolution were used in EMG equipment and quite recently Walters et al. (2013) found a portable electromyograph with an 8 bit ADC to be a valid solution for field measurements. A conclusion may be drawn that in terms of range, the contemporary low-cost hardware has potential to render the EMG amplitude truthfully.

In the case of EMG frequency characteristics, capturing the up to 500 Hz proportion of the raw EMG requires a sampling rate of ≥1000 Hz. The sampling rate is controlled by the microcontroller and some of the low-cost single-board microcontrollers are capable of acquiring the EMG from a few channels simultaneously (table 4). The number of the single-board microcontroller's channels that can be used for raw EMG acquisition depends on the sampling frequency, thus one way to increase the number of channels is to narrow down the EMG spectrum sampled. According to Merletti & Torino (2014), it is acceptable to narrow the upper portion of the EMG to 350 Hz. Even with this approach the contemporary low-cost electromyographs are no match to the commercial devices, which allow sampling with 16 or 32 channels. Meanwhile the number of analogue channels on a contemporary low-cost microcontroller board ranges from four to sixteen and only a proportion of the channels can be used with sample rates necessary for EMG acquisition (see table 4, p. 31). The narrowing can be also applied to the lower portion of the EMG. However, this has nothing do with the sampling rate and is meant to reduce the signal artifacts (Clancy et al., 2002). In this case, a portion of the spectrum which lies below 10-20 Hz is disregarded (see table 1, p. 14). It is evident from table 2 (p. 25) that only the BITalino v.151015 has the hardware filter characteristics that satisfy the criteria of raw EMG acquisition in terms of the EMG spectrum. Meanwhile, the low-pass filter with the cut-off frequency set to 40 Hz of the Olimex Shield EKG/EMG neglects the majority of the EMG spectrum and using this amplifier is therefore not recommended. The remaining two bioamplifiers, which output raw EMG, do not have filters at all; this is definitely preferred to having improper filters as data may be also filtered digitally post-recoding.

In contrast to the raw EMG acquisition, the demands for the acquisition of linear envelope are far more achievable by the contemporary low-cost single-board microcontrollers. According to Merletti & Torino (2014), sampling rates between 50 to 100 Hz are sufficient while acquiring the linear envelope of EMG. Note, that the majority of the bioamplifiers listed in table 2 output raw EMG; this allows conventional fatigue assessment, as raw EMG is necessary to analyse shifts in the EMG spectrum (Merletti et al., 1991). However, obtaining V_{RMS} or linear envelope from raw EMG requires processing of the stored data. The two bioamplifiers from Advancer Technologies, LLC (Muscle Sensor v3 and Myoware) output the linear envelope of detected myoelectric activity. Although recorded linear envelope does not allow fatigue assessment by analysing the shifts in the EMG spectrum, fatigue assessment may still be possible, as fatigue is known to increase the EMG amplitude of the same effort (Cifrek et al., 2009; Merletti et al., 1991). EMG amplitude by itself is sufficient for three out of the four main uses of electromyography in ergonomics (Marras, 1990). In addition, linear envelope (or EMG amplitude) is preferred to raw EMG in control or feedback applications. Thus a conclusion may be drawn that the contemporary low-cost hardware has potential to detect and record myoelectric activity. The remaining two tasks, storing and/or displaying the myoelectric activity, may be better solved outside the realm of low-cost hardware by using a PC (**I**). This may still require a DIY approach; however, in this case the focus shifts from efficiency to effectiveness.

Considerations for practice

Efficiency considers time, human effort, and money in order to decide whether using the DIY approach is reasonable. One of the main differences between the DIY and commercial solutions lies in the software options to communicate with the device, process, visualise, and store data. In the realm of low-cost hardware, the software options are somewhat limited. BITalino is accompanied by freeware 'OpenSignals (r)evolution' (PLUX wireless biosignals S.A., Portugal), which allows communication with the device, storing data, and data visualisation. Data processing for further analysis requires a paid add-on. Several options to process EMG are available when operating without specially-tailored software – Labview, Matlab or even Microsoft Excel. Microsoft Excel is usually installed on the PCs as part of common office software. Meanwhile LabVIEW and Matlab are development environments which require purchasing a somewhat expensive license. The advantages of development environments include the availability of toolboxes/toolkits for EMG processing and the fact that data processing can be easily automated. While Microsoft Excel offers significantly less automated data processing, all aspects of EMG processing can be achieved by following the tutorial of Robbins (2014). A key conclusion from the discussion above is the apparent presence of a trade-off between the financial and temporal expenditures (**I**).

The trade-off is especially clear in the case of Microsoft Excel, which is a basic and universal data processing tool and does not require additional financial expenditure. However, processing EMG with Microsoft Excel (**I**, **II**), is a time-consuming task. By rough estimates, extracting a single contraction for analysis will take about 15 s with commercial software, about 60 s with Matlab and about 120 s with Microsoft Excel. In a simple case, e.g. 10 subjects, one muscle, three contractions (x_i, x_{ref}, x_{min}) and two repeats, the analysis would take >120 min with Microsoft Excel and >15 min with commercial software. The main temporal advantage of commercial software is the 'point and click' procedure, while Matlab and Microsoft Excel to an extent rely on keyboard data entry. Microsoft Excel might be an appropriate tool for processing hardware filtered EMG amplitude, but it is not convenient to process large amounts of data (**I**). The inconvenience and temporal expenditure of data processing increase substantially when studying complex tasks and multiple muscles. The latter is common among EMG studies in industrial settings where the focus is on the prime movers of fingers and wrist (Kadefors et al., 1993). Therefore, data processing with Microsoft Excel might be better suited for situations where complexity is low and temporal requirements are less strict, e.g. students gaining their first experience about EMG in higher education. The DYI data processing might even help relieve some of the concerns about EMG misuses raised by Cavanagh (1974) and De Luca (1997). After all, it is vital for an ergonomist to interpret EMG related research (Ankrum, 2000) and the commercial 'black box' software may avert one from obtaining the necessary insight in the first place (Robbins, 2014). Fundamental knowledge about EMG processing is also necessary when using Matlab, as one has to write a script which later can be easily rerun and thus to a degree allows to automate data processing. Matlab grants easy handling of data from simple single contractions to more complex muscular activity (as in **III**). Several authors (Chowdhury & Nimbarte, 2017; Hoozemans & van Dieën, 2005; Walters et al., 2013) have used Matlab to process EMG, thus using Matlab is not an uncommon practice, especially in the case of custom hardware or innovative research goals. A conclusion may be drawn that several options are available to process EMG data that is acquired with the contemporary low-cost hardware; however, achieving the desired end result does not depend only on the efficiency but also on the expertise and skills of the users.

While there are fundamental skills or knowledge necessary to work with electromyography, such as electrophysiology, muscle functions and muscle mechanics (Cavanagh, 1974), there are also skills unique to the DIY approach. Basic mechanical or electrical engineering knowledge, such as soldering (**I**) or computer-aided design to prepare the protective cases (Reinvee & Mrugalska, 2019) may be necessary to use the low-cost hardware. In addition, knowledge about computer engineering is necessary to program the microcontroller board or write Matlab or LabVIEW scripts. Optionally, it is possible to reduce the requirements to the skillset, as BITalino incorporates sensors with standard USB plugs, free 3D printing models for enclosures, and paid add-ons for EMG analysis. Therefore, the trade-off between the willingness to spend financial resources and the required skillset exists also in the realm of contemporary low-cost electromyography equipment. In the end, irrespective to the specific low-cost hardware, the overall cost of the low-cost electromyography equipment is only a fraction of a commercial system. This makes low-cost hardware especially attractive for: i) the users in growing economies (Stojanovic et al., 2015), ii) situations involving a risk of damaging the equipment (Walters et al., 2013) and iii) higher education or life-long learning (**I**, **II**). Having multiple low-cost electromyographs available in the classroom allows the students to learn by doing (**I**) which is deemed to be superior to reading or listening by the approaches like Dale's cone of experience or the 70:20:10 framework. This allows challenging the order in which the necessary skills or prerequisites to work with electromyography are obtained, as electromyography may be also used to identify the functions of muscles or demonstrate fundamental aspects of electrophysiology. This approach may be especially beneficial for the kinaesthetic learner of Fleming's (1995) VARK model. Irrespective to the learning style or the level of expertise, encouraging one to use low-cost electromyography equipment requires confidence in the equipment's reliability.

There are several ways to make sure that the low-cost electromyography equipment works as intended. One could compare the performance of a low-cost system against an established gold-standard, i.e. to a commercial electromyograph (**II**, Batista et al., 2019; Walters et al., 2013). Alternatively, one could take advantage of the fact that the microcontroller boards can be easily and simultaneously interfaced with many additional types of sensors such as electronic dynamometers, load cells, and accelerometers (**I**, **III**). Firstly, this is an attractive feature by itself, as an isolated EMG hardly allows meaningful interpretation (Cavanagh, 1974). Secondly, the relationship between an electromyogram and a dynamogram is expected to be at least partially linear, thus obtaining such a relationship would indicate that the electromyography equipment adequately reflects the real situation. Figure 5 indicates a fairly linear relationship between EMG amplitude and muscular effort. Some scatter of data may be observed when the effort exceeded 50%MVE, however, this in line with Solomonow, Baratta, Zhou, Shoji, & D'Ambrosia (1987) and associated with increased firing rates of motor units (Milner-Brown et al., 1973). In addition, the absence of statistically significant differences when low-cost electromyography equipment was compared to a commercial electromyograph (figure 6) also indicates that DIY electromyography is a valid approach at least in the case of submaximal contractions. At the same time it is reasonable to maintain healthy scepticism relating to the lack of statistically significant differences between the results of DIY and commercial equipment, as the sample size n =10, might not be high enough to avoid type II error, i.e. non-rejection of false negative result. However, as electromyography studies require substantial resources, it is common to conduct the studies with 'small' sample sizes, which seldom exceed 15 subjects (Mathiassen, Burdorf, & van der Beek, 2002).

As already mentioned in the discussion above, the gain factor of the commercially available fixed gain bioamplifiers is not ideal and could cause the amplifier to saturate and distort data. However, one can hardly find complaints about improper gain in the literature. There may be several reasons why this is not extensively addressed in the literature: i) maximum voluntary efforts are more likely to cause amplifier saturation than submaximal efforts, ii) EMG amplitude depends on several factors (Cram & Kasman, 2010), thus saturation might only occur in the case of some muscles, iii) electromyography is used in many different areas, namely sport, occupational health and safety, electromyography-controlled prostheses or robotic exoskeletons, each introducing its specifics (Guo, Sandsjö, Ortiz-Catalan, & Skrifvars, 2019). It is reasonable to expect that most of the areas using electromyography are interested in submaximal efforts, while MVE is often used to normalise the EMG amplitude. Walters et al. (2013) tested a low-cost apparatus with parameters similar to BITalino and found that saturation occurred among some subjects in the case of *m. vastus medialis* but not in the case of *m. tibialis anterior* nor in the case of *m. biceps brachii*. Some authors have been faced with problems opposite to saturation. Gussev et al. (2018)

used BITalino and observed a visible but relatively small response in the absolute EMG amplitude (in μV) of *m. erector spinae* when studying trunk flexion manoeuvres. The authors concluded that their application would benefit either from a bioamplifier with a higher gain or from an ADC with a higher resolution. Considering that both the subjects' and muscles' parameters affect the EMG amplitude (Cram & Kasman, 2010), it is evident that when contemporary low-cost apparatus are used then a major electromyography study should follow a pre-study, where the latter has a major role determining: i) whether the muscles can be studied with the fixed gain amplifiers; ii) how to set the adjustable gain amplifier; iii) how to normalise the EMG amplitude; or iv) if the resolution is sufficient to distinguish changes in muscular load or effort. It may be observed from figure 7 (p. 33), that nEMG reflects changes in effort. As expected, the changes are statistically significant when comparing effort required to simply hold the power tool and the increased effort required to hold the power tool with the motor running. An additional effort related to cutting the material does not appear to be statistically significant. This is partially due to the inter-participant variance in anthropometrical parameters and task execution. In addition, one may doubt whether there is supposed to be a significant increase in the muscular effort that should be reflected in EMG amplitude. It is the abrasive disk that is supposed to do the work, not the muscular system. This doubt is supported by other objective methods used in the in paper **III** (for details see p. 88). Thus paper **III** (but also **I** and **II**) has shown that there are muscles, tasks and subjects that can be studied with the means of DIY low-cost electromyography hardware and software resources.

Low-cost applications

It may be gathered from the discussion above that utilising the DIY low-cost hardware and software resources involves some uncertainty which can hardly be resolved prior to actual measurements. Although suitable gain and range may be determined in a pre-study, this indicates that the low-cost resources are usable in the case of low temporal demands, or when financial demands outweigh temporal demands, e.g. when there is a risk for damaging the equipment (Walters et al., 2013) or a lack of financial resources restricts the access to an expensive commercial electromyograph (Supuk et al., 2014). One environment where both of these considerations apply is academia. For an ergonomist the interpretation of EMG is crucial (Ankrum, 2000). It is believed that mastering this knowledge requires hands-on supervised learning and the possibility of the latter depends on access to the apparatus. The literature review (p. 13) highlighted multiple electrode-related procedural issues which do not concern the apparatus but affect EMG data. These issues require practical training and will wear the apparatus. Heywood et al. (2018) mentioned anecdotally that low-cost electromyographs may be quite robust and the same may be said about the solutions used in **I** and **II**. This encourages using the DIY low-cost apparatus in higher education to learn the EMG acquisition procedures and possibly even for research. Some authors have used low-cost bioamplifiers with more expensive 14 or 16 bit ADCs (Heywood et al., 2018; Lu et al., 2019; Supuk et al., 2014). This somewhat increases the cost; however, these data acquisition devices may be interfaced with several different sensors, thus offering a flexible and budget-friendly research setup.

In contrast to research, the requirements for electromyography feedback are less strict. DIY low-cost apparatus can be used to detect muscles' on-off patterns and certainly for semi-quantitative assessment of relative effort (**II**). The linearity between nEMG and relative grip force that was demonstrated in figure 5 encourages the use of the DIY low-cost apparatus in electromyography feedback devices. Peper et al. (2003) report observations that office workers lack awareness about their muscle tension. Electromyography feedback devices might be used to teach workers to properly relax their muscles and to improve posture. Having this particular goal in mind, the user needs to be aware which muscles need to be inactive in the desired posture and where to attach the electrodes on the muscle of interest. Some of the amplifiers (e.g. Myoware) are designed in a way which allows reducing electrode-related misuses (i.e. electrode sockets are recessed to the printed circuit board). The precise knowledge of anatomy is probably not the limiting factor for the usage of the electromyography feedback device. This knowledge may be taught or self-learned via anatomy atlases, or even by trial and error using the feedback device itself.

The two-channel electromyography feedback device prototype constructed in this thesis cost about 120 euros to make, which qualifies it as a low-cost device. On the one hand, reducing EMG-channels to only one would reduce the cost to 70 euros. On the other hand, the two-channel prototype allows instant comparison of muscular load and more understandable demonstration of the hand functions (e.g. flexion-extension)

CONCLUSIONS

The analysis of the technical characteristics of contemporary low-cost electromyographs and their compliance with the hardware requirements showed that the characteristics of the bioamplifiers studied in the thesis to a large extent meet the standards described in the literature. Meanwhile, the measurement range or the dynamic range of available single board microcontrollers may hinder the effectiveness of low-cost do it yourself (DIY) electromyographs. There are two strategies to overcome this issue: either using a bioamplifier with adjustable gain or normalising the electromyograms far below the maximum voluntary effort. Both strategies require time and effort, either to determine an appropriate reference effort for amplitude normalisation or to find a suitable gain value, thus reducing the efficiency.

The evaluation of the quality of contemporary low-cost electromyographs showed that the tested low-cost DIY electromyographs were effective enough to study the selected forearm muscles. Their performance was found to be satisfactory and the results comparable to the results obtained with a commercial electromyograph (**II**). However, achieving a comparable result required more work from the investigator and a higher level of expertise in the case of DIY low-cost electromyographs than in the case of the commercial electromyograph.

The low-cost DIY electromyographs can be applied in the ergonomic assessment of hand tools. A low-cost electromyograph interfaced with an accelerometer was successfully used to assess the ergonomic quality of an angle grinder (**III**) and a prototype of a DIY low-cost electromyography feedback device capable of demonstrating hand function was created.

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KOKKUVÕTE

MADALA MAKSUMUSEGA ELEKTROMÜOGRAAFIDE RAKENDATAVUS ERGONOOMIKALISES HINDAMISES

Luu- ja lihaskonna ülekoormushaigused on globaalne tööjõu efektiivset kasutamist pärssiv probleem, mis avaldub sissetulekute vähenemisena nii üksikisiku, ettevõtte kui ka riigi tasandil. Seetõttu on osapooltel huvi tööga seotud luu- ja lihaskonna ülekoormushaiguste ennetamiseks. Ennetus toimub reeglina ettevõtte tasandil, kus töökeskkonna riskianalüüsi käigus selgitatakse välja töökeskkonna ohutegurid ja seejärel rakendatakse meetmed ohutegurite mõju vähendamiseks. Seega sõltub ennetus riskianalüüsis kasutatavate meetodite asjakohasusest.

Luu- ja lihaskonna ülekoormushaiguste tekkega seotud töökeskkonna füsioloogilise ohuteguri hindamiseks kasutatakse peamiselt ankeetmeetodeid. Ankeetmeetodite oluliseks metoodiliseks puuduseks on nende subjektiivsus, mistõttu tuleb võimalusel eelistada objektiivseid meetodeid ehk mõõtmisi. Sageli piirab mõõtmiste kasutamist seadmete kättesaadavus ning seda peamiselt nende maksumuse tõttu.

Viimasel kümnendil on oluliselt laienenud füsioloogiliste parameetrite mõõtmise võimalused, vähenenud on seadmete mõõtmed ja maksumus. Saadaval on komponendid, mida saab kasutada mõõtmis- ja juhtimisseadmete valmistamiseks *isetegemise* põhimõttel. Need komponendid on mõeldud kasutamiseks mitmetes valdkondades, näiteks telemeditsiin, inimese-arvuti interaktsioon või inimese jõudlust parendavate turvismasinate valmistamine. Eeltoodud valdkondade ja töökeskkonna riskianalüüsi eesmärgid on erinevad ning seetõttu pole teada, kas nende komponentide abil on võimalik luua seadmeid, mida saab kasutada luu- ja lihaskonna ülekoormushaiguste riski hindamisel. Väitekiri keskendub ühe füsioloogilise parameetri mõõtmise võimalusele – lihase biopotentsiaalide mõõtmisele naha pinnalt ehk pinnaelektromüograafiale.

Väitekirja eesmärgiks oli hinnata nüüdisaegse madala maksumusega elektromüograafi kasutamise võimalusi luu- ja lihaskonna ülekoormushaiguste ennetamisega tegeleva rakendusteaduse, ergonoomika valdkonnas. Eesmärgi saavutamiseks: 1) analüüsiti nüüdisaegsete madala maksumusega elektromüograafide tehniliste omaduste vastavus riistavarale esitatud kriteeriumitele, 2) hinnati madala maksumusega elektromüograafia kasutatavust labori- ja välitingimustes, 3) loodi ja katsetati ergonoomika otstarbelisi madala maksumusega elektromüograafia rakendusi.

Väitekirja materjalide alusel avaldati kolm publikatsiooni eelretsenseeritavas ajakirjas ja kogumikes. Esimeses publikatsioonis kaardistati madala maksumusega elektromüograafi kasutamise võimalused. Publikatsioonis esitatud elektromüograafia võimendite parameetrite ning kriteeriumite võrdlus on väitekirjas esitatud laiendatud kujul ja publikatsioonis välja pakutud seadmete koostamise ning programmeerimise skeemid on väitekirjas esitatud muudetud kujul. Teises publikatsioonis võrreldi laboriuuringus isetegemise põhimõttel koostatud elektromüograafe kommertsseadmega ja kolmandas publikatsioonis kasutati väliuuringus *isetegemise* põhimõttel koostatud elektromüograafi käsitööriista ergonoomikalises hindamises.

Tehniliste parameetrite analüüsis vaadeldi võimendite mürasummutamisega seotud omadusi, mikrokontrollerite arendusplaatide andmehõive seotud omadusi ja võimendite ning mikrokontrollerite arendusplaatide omavahel seotud mõõtmispiirkonda ja eraldust puudutavaid parameetreid. Selgus, et viiest uuritud võimendist neli vastas elektromüograafia standarditele ISEK ja SENIAM, kuid mõningaid piiranguid võib seada uuritud võimendite kasutamine koos nüüdisaegsete madala maksumusega mikrokontrollerite arendusplaatidega. Kui võimendi võimendusteguri ja mikrokontrolleri arendusplaadi mõõtepiirkonna väärtuste suhe pole optimaalne, siis ei pruugi olla võimalik saada moonutusteta elektromüogramme iga inimese, lihase või tööülesande mõõtmisel. Kuna konkreetses olukorras nõuab seadme korrektses toimivuses veendumine täiendavat ajakulu, võib seda lugeda puuduseks. Kommertsseadmete eeliseks on seadmega kaasnev ja spetsialisti tööd kiirendav mõõtmistulemuste analüüsi tarkvara. Andmeid on võimalik analüüsida ka Microsoft Exceli abil, kuid see on aeganõudev ja eeldab kasutajalt suuremat pädevust kui kommertstarkvara kasutamine. Kõrgemad nõuded seadme kasutaja pädevustele tulenevad ka vajadusest veenduda seadme korrektses toimivuses ning vajadusest luua ja valmistada seadet välismõjude eest kaitsev korpus.

Alla 150 euro maksva *isetegemise* põhimõttel koostatud elektromüograafi eelised kuni kümneid tuhandeid maksva kommertsseadmete ees tulenevad avatud lähtekoodist, seadme kohandatavusest ja maksumusest. Seadme madal maksumus teeb elektromüograafia kättesaadavaks ega pärsi seadme kasutamist kulumise või hävinemise ohu tõttu. Kohandatavus ja avatud lähtekood võimaldavad seadmesse integreerida mitmeid andureid ja täitureid, mille abil on spetsialistil elektromüogramme lihtsam interpreteerida või luua elektromüograafial põhinevaid tagasisideseadmeid. Elektromüogrammide õigest interpreteerimisest sõltub riskihindamise kvaliteet. Tagasisideseadmeid saab spetsialist kasutada töötajale optimaalsete töövõtete ja -asendite õpetamiseks.

I

ORIGINAL PUBLICATIONS

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Overview of Contemporary Low-cost sEMG Hardware for Applications in Human Factors and Ergonomics

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For decades, surface electromyography (sEMG) has been one of the essential methods of Human Factors and Ergonomics. Although capturing sEMG data is relatively easy, proper interpretation of acquired data is possible only with sufficient background in electrophysiology, muscle mechanics and muscle functions. As the last decade has made biosignal acquisition more accessible, the current overview discusses the intrinsic properties of contemporary low-cost sEMG acquisition systems and proposes applications of sEMG for Human Factors and Ergonomics.

INTRODUCTION

Electromyography (EMG) is a field specializing in the use of electronic devices to measure the bioelectrical activity of muscles and analyze the data. In Ergonomics, the noninvasive assessment of neuromuscular activity e.g. surface electromyography (sEMG) is used. Marras (1990) lists four primary applications of sEMG in Ergonomics: (i) detection of muscle activity and inactivity (i.e. on-off duration); (ii) comparison of relative muscle activity levels under various tasks as indication of muscle effort; (iii) quantitative assessment of muscular force, however, this is only possible in very strict cases (e.g. isometric or isokinetic contractions); and (iv) fatigue assessment by analysis of spectral components of raw sEMG signal. In addition, some authors have proposed assessment of muscular strength by sEMG spectrum (Jung, 1987) and simultaneous analysis of sEMG spectrum and amplitude for muscle strain and fatigue assessment (Luttmann, Jäger & Laurig, 2000).

Although sEMG is considered one of the essential tools in Ergonomics, only one third of certified professional ergonomists reported ever using electromyography (Dempsey, McGorry & Maynard, 2004). Surprisingly only 8.7% of ergonomists who had never used sEMG reported it to be too expensive. However, 23.7% noted that electromyography is not available for them.

More than 40 years ago, Cavanagh (1974) emphasized that surface electromyograms are simple enough to be obtained by anyone with the right equipment and a few minutes of instruction, but analysis of acquired data without the basic understanding of fundamental electrophysiology, muscle function and mechanics is prone to misinterpretation. The increased availability and the reduced cost of various electronic devices during the last decade might have intensified the problem.

Therefore, the goal of this overview is to discuss contemporary low-cost instrumentation for recording electromyograms and the ways to utilize such devices in Human Factors and Ergonomics.

INTRINSIC PROPERTIES OF LOW-COST SEMG ACQUISITION SYSTEMS

Hardware considerations

In order to successfully capture the sEMG signal, the following procedures are needed (Figure 1): sensing of the phenomenon of bioelectric activity, conditioning of the signal (amplifying, filtering, analog to digital conversion) and finally, displaying or storing the acquired data. The obvious choice for data display or storage is the personal computer, as it is usually referred to as 'universal lab equipment'. The remaining tasks can be divided between a single-board microcontroller and a shield. The latter is defined as an electronic board, which can be connected to a microcontroller in order to add functionality.

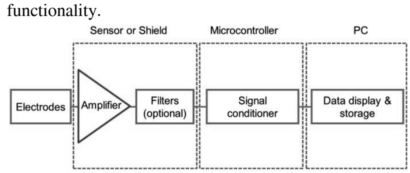

Figure 1. Elements of the sEMG acquisition system

The most popular single-board microcontroller, Arduino is already proposed as a low-cost tool for psychological and neurophysiological labs (D'Ausilio, 2012) and it has been proven to be capable of capturing and reproducing the EMG signal (de Moura, Ozelim, & Soares, 2014). The main advantages of Arduino boards are the low cost (starting ~$25), small dimensions and easy to learn programming language. In addition, beside Arduino, there are several companies on the market, which are manufacturing generic or their own Arduino-compatible microcontrollers. However, some of the critics claim that at least some of the low-cost microcontrollers do not conform to the specific needs of bio-signal acquisition (Guerreiro, Lourenço, Silva & Fred, 2014). In general, the contemporary microcontrollers have at least six analog channels with a measurement rage of 0 to 3.3 V or 0 to 5 V. Meanwhile, the EMG signal is bipolar by its nature. Thus, the EMG sensors or shields need to imply DC offset and set the baseline to half of the microcontrollers' measurement range. Most of the microcontrollers have 10 bit analog to digital converters (ADC-s), which means that continuous signal is transformed to 2^{10} discrete values and sampling resolution is either 3.2 mV (3.3 V microcontrollers) or 4.9 mV (5 V microcontrollers). Some of the 3.3 V microcontrollers (e.g. Arduino Due) have a 12 bit ADC and therefore have 0.8 mV sampling resolution. Meanwhile, peak-to-peak amplitude of the sEMG signal is known to reach 5 mV (Kamen & Gabriel, 2010), thus adequate amplification is needed in order to avoid either missing or clipping of the signal. The shield gain factor should not exceed 660 in the case of 3.3 V microcontrollers and 1000 in the case of 5 V microcontrollers.

In order to ensure electrical safety, the microcontroller needs to be insulated from the power line; this can be achieved with wireless data transfer or an USB isolator.

An in-depth internet search was conducted in order to find microcontroller-compatible sEMG sensors/shields. The results needed to satisfy the following criteria: 1) cost below $150; 2) the product is sold either by manufacturers' or distributors' web-store and 3) absence of negative feedback in the web store comments section. Solutions which included a significant amount of do-it-yourself approach were excluded. Four products satisfied the criteria: (i) MyoWare (Advancer Technologies, Raleigh, USA); (ii) 'Olimex Shield EKG/EMG' (OLIMEX Ltd, Plovdiv, Bulgaria); (iii) BITalino (PLUX wireless biosignals S.A., Portugal) and (iv) FlexVolt (Flexvolt Biosensor, Lebanon, USA). The main properties of the shields are summarized in table 1.

Table 1. Properties of sEMG shields

Shield or sensor	Channels (ch)	Cost	Signal type	Filters	Gain
BITalino EMG sensor	1	$80	raw	Band pass 10-400 Hz	1000
FlexVolt Shield	2 or 4	$60/ $95	raw	None	2336
Olimex Shield EKG/EMG	1	$53	raw	Low pass fc = 40 Hz	adjustable
MyoWare	1	$38	raw / rectified	none for raw signal	adjustable

It is commonly accepted that the prominent part of the sEMG spectrum range lies between 5–500 Hz, however, frequency range 10 to 350 Hz is preferred by ISEK standards for reporting EMG data (Merletti & Torino, 1999). The sEMG shield does not necessary need to contain hardware filters, as the researcher may add digital filters in data processing software (e.g in Matlab or LabView). Moreover, inappropriate hardware filters are impossible to remove during data processing. Filtering is still a necessity as the lower range of the sEMG spectrum is known to contain motion or electrocardiography (ECG) artifacts. Motion artifacts are reported to lie below 10 Hz (Ankrum, 2000; Solomonow, 2000) or 20 Hz (Clancy, Morin, & Merletti, 2002; De Luca, Gilmore, Kuznetsov & Roy, 2010), the ECG artifact is reported to lie below 35 Hz (Christov & Daskalov, 1999). Thus, the primary application of Olimex Shield EKG/EMG is electrocardiography measurement, as it excludes most of the sEMG spectrum and is prone to pick up movement artifacts.

It has been suggested to locate the amplifiers close to the pick-up area (Marras, 1990); this concept is adopted in the MyoWare sensor (Figure 2). The sEMG electrodes are attached to the board with a constant inter-electrode distance of 3.0 cm.

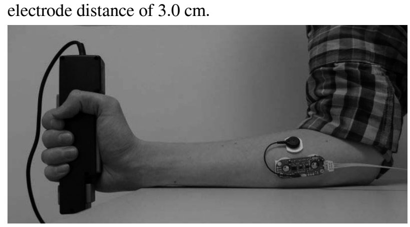

Figure 2. MyoWare sensor attached to forearm flexor muscle group, during grip force measurement.

The user can change the output of the MyoWare sensor by a hardware switch – either set it to capture raw sEMG signal or to smoothed-rectified signal, e.g the absolute value of the signal. The latter can be directly used in control or feedback applications, meanwhile raw sEMG data requires processing (e.g FTT or RMS).

The gain factors of the BITalino sensor and the FlexVolt seem to exceed the optimal values. However, preliminary tests allow to conclude that the gain factor is not problematic, at least when acquiring sEMG data from forearm flexors and extensors.

Software considerations

The main disadvantage of low-cost sEMG tends to be a lack of appropriate data processing software. There also seems to be a tradeoff between the cost and time expenditure. Some researchers have reported using Microsoft Excel (Burden & Bartlett, 1999; Reinvee, Vaas, Ereline & Pääsuke, 2015). Microsoft Excel is a universal data processing tool, which can be used when processing adequately hardware-filtered sEMG amplitude, but it is not convenient to process large amounts of data. The Iowa EMG Analysis Program (IEAP) (Fethke, Anton, Fuller & Cook, 2004) seems to be a more user friendly alternative; however, more than decade after the publication, we were unable to find any trace of the IEAP. Meanwhile, one can find LabView or Matlab toolboxes form the internet for every aspect of EMG data processing. These options can be tailored to the users' needs; however, the user should be familiar with the software, also, one might find the cost of software license unreasonable high for infrequent use. The paid 'Electromyography analysis' add-on for 'OpenSignals (r)evolution' freeware (PLUX wireless biosignals S.A.) seems to include most of the necessary features.

LOW-COST SEMG APPLICATIONS IN HFE

Several authors stress the importance of researchers' and practitioners' knowledge and familiarity with the sEMG capture and data interpretation process (De Luca, 1997; Ankrum, 2000; Clarys, 2000; Marras, 2000; Solomonow, 2000). Meanwhile, in a survey of industrial ergonomics tool use, only 1.6% of certified professional ergonomists reported not being familiar with electromyography (Dempsey, McGorry, & Maynard 2004).

Electromyography is known to be an expensive tool, therefore it is expected that reducing the price of instrumentation would have an impact on the utilization of sEMG in HFE education. Having more than one instrument in the classroom allows to increase the student engagement and implement the concept of learning by doing. Although the question of the exact placement of sEMG electrodes is at least to some extent debatable (Hermens, Freriks, Disselhorst-Klug & Rau 2000; Hoozemans, Loos, Wilms, & Dieën, 2006; Mesin, Merletti, & Rainoldi, 2009; Takala, & Toivonen 2013), it is strongly recommended that an experienced instructor have sufficient time to assist all the students.

One of the main advantages of the low-cost sEMG acquisition systems is the microcontroller's compatibility with many other sensors (electronic dynamometers, load cells or accelerometers). Thus, the relation between the surface electromyogram and muscular force during isometric contractions (Milner-Brown & Stein, 1975) and its linear or non-linear properties (Solomonow, Baratta, Zhou, Shoji, & D'Ambrosia, 1987) can easily be studied in laboratory exercises. Example results of such laboratory exercises are shown in Figures 3 and 4. The data acquisition followed the procedure of the Caldwell regimen (Caldwell et al., 1974). The MyoWare sensor was attached to *m. flexor digitorum superficialis* and the sensor was connected to an Arduino Leonardo microcontroller. An electronic dynamometer (Vernier Software & Technology, Beaverton, USA) was also connected to the microcontroller, which allowed simultaneous data capture.

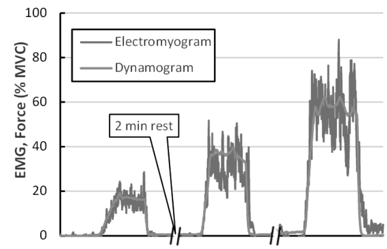

Figure 3. Grip force dynamograms and electromyograms of *m. flexor digitorum superficialis* during submaximal contractions (own source, unpublished data).

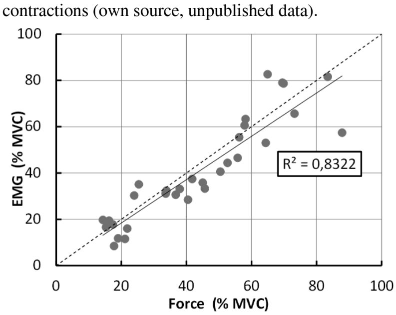

Figure 4. Relation between the *m. flexor digitorum superficialis* electromyogram and grip force, unpublished data of six males.

Based on simplicity, either the MyoWare sEMG sensor (pros: budget friendly, customizable; cons: requires some soldering) or the BITalino system (pros: software, customer service, plug and play hardware; cons: price) can be recommended for educational purposes.

SEMG has also been proposed as an onsite training tool for musculoskeletal disorders. Peper et al., (2003) carried out intervention programs in office settings and noted that workers have low awareness of their muscle strain and suggest that a training specific to particular tasks and equipment needs to be conducted at the job site. The sEMG biofeedback devices have also been reported to increase audience engagement while explaining ergonomics principles: "The students especially seemed to enjoy being hooked up to the Pocket Ergometer, which converts the electrical activity associated with muscle use into sound. Students were able to hear the difference between neutral and awkward postures and between wearing a backpack with one strap or with two" ("Puget Sound Chapter Outreach Activities", 2005). The 'Pocket Ergometer' (Biomechanics Corp. of America, USA) is a sEMG feedback device, which measures bioelectric activity and gives visual or auditory feedback. We propose that a similar device can be assembled from a MyoWare sensor, an Arduino compatible microcontroller, colored LEDs, a piezo buzzer and an LCD display with less than $100. The proposed device needs to perform the tasks, which are shown in Figure 5.

0. Provide sEMG feedback
Plan 0: Do 1 then 2 (if necessary) then 3 & 4 then repeat 3 & 4
- 1. Start device
  Plan 1: Do 1.1 then 1.2 & 1.3
  - 1.1. Set default values for rest and MVC sEMG
  - 1.2. Set audio feedback pitch range
  - 1.3. Read calibration button state
- 2. Calibrate
  Plan 2: Do 2.1 then 2.1 then 2.2 then 2.3
  - 2.1. Blink LEDs to indicate start of calibration
  - 2.2. Initialize calibration sequence
    Plan 2.2: For 10 seconds repeat 2.2.1–2.2.4
    - 2.2.1. Take 10 readings
    - 2.2.2. Calculate average
    - 2.2.3. If average < default rest value then store new rest value
    - 2.2.4. If average > default MVC value then store new MVC value
  - 2.3. Blink LEDs to indicate end of calibration
- 3. Measure
  Plan 3: Do 3.1 & 3.2
  - 3.1. Take 10 readings
  - 3.2. Calculate average
- 4. Provide feedback
  Plan 4: Do 4.1 then 4.2 or 4.3 then 4.4
  - 4.1. Read feedback button state
  - 4.2. Provide visual feedback
    Plan 4.2. Do 4.2.1 then 4.2.2 and /or 4.2.3
    - 4.2.1.Scale calculated average
    - 4.2.2.Write numeric results on LCD display
    - 4.2.3.Turn on specific number of LEDs
  - 4.3. Provide audio feedback
    Plan 4.3. Do 4.3.1 then 4.3.2
    - 4.3.1.Scale calculated average
    - 4.3.2.Play a tone in the pitch range
  - 4.4. Wait 10 ms

Figure 5. Hierarchical list for the software of the sEMG feedback device.

It is possible to write the necessary microcontroller code solely on the basis of built-in examples included in the Arduino software. For a demonstration video refer to: https://www.youtube.com/watch?v=2l5GvTqeB1E

CONCLUDING REMARKS

It should be noted that hardware properties or availability by itself do not have an impact on the use or abuse of sEMG. The practitioners are usually bound by the code of ethics of their professional society to limit their practice to the areas they maintain sufficient competence in. Therefore, one should have basic understanding of instrumentation, anatomy and physiology in order to successfully use contemporary low-cost sEMG acquisition systems. In the case of education, it is a matter of debate whether an understanding of instrumentation, anatomy and physiology is a prerequisite to the use sEMG or might low-cost sEMG devices be utilized to learn the necessary concepts.

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Applicability of affordable sEMG in ergonomics practice

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Abstract

This study was conducted in order to assess the potential of low-cost surface electromyography (sEMG) measurement systems to be applied in ergonomics practice on the example of provided relative grip force estimates. Two measurement configurations, with a total cost below $100, were compared with a commercially available electromyograph. It was hypothesized that Arduino based do-it-yourself measurement systems do not perform significantly worse than commercially available electromyograph. Ten participants' normalized sEMG activity of the m. flexor digitorum superficialis were compared in the case of three submaximal isometric hand grip force levels (25%, 50%, 75% of maximum voluntary contraction). Normalized sEMG activity, measured with different systems, was not statistically different. It was concluded that low-cost measurement systems have the potential to be used in order to detect muscle activation-deactivation patterns or at least the semi-quantitative assessment of grip force; however, the results of this study are limited to isometric contractions.

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Keywords: Low-cost EMG; Arduino in ergonomics; Isometric gripforce assessment

1. Introduction

Musculoskeletal disorders (MSD-s) have been a key topic in ergonomic research for decades. Putz-Anderson [1] listed the use of excessive force, highly repetitive work process, awkward postures and inadequate rest as the main causative factors for MSD-s, but more recent reviews have listed up to 70 [2] potential causative factors. Obviously it is impossible to consider such an amount of factors in experimental studies. Most of these factors can be assessed

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with self-administered questionnaires and the prevalence of some factors can be detected by observations but force evaluation still requires apparatus. Often Borg's perceived exertion scale [3] is used instead of an apparatus; however, some occupational studies report up to 50 % overestimation in self-reported force exertions [4].

The main challenge of force measurement in occupational settings is the representativeness of the results, thus the apparatus should not alter the task or require interference in the work process. Both force sensing resistors (FSR-s) [5] and surface electromyography (sEMG) [6] will satisfy such criteria.

In a review Hägg et al. [7] lists three major sEMG applications in ergonomics practice: 1) detecting activation patterns of muscles; 2) estimation of muscular force and 3) fatigue assessment. It is obvious that the second and third applications are impossible without the first. Muscular force and muscle fatigue assessments with sEMG do not necessary need to be performed simultaneously. Often the EMG power spectrum is used in fatigue analysis, where changes in median or mean frequency are used as fatigue indices [8–10]; however, critics of this approach suggest joint analysis of the EMG spectrum and amplitude, i.e. the JASA method [11, 12].

Several authors have explored the possibility of estimating grip force exertion via EMG amplitude [13–20]. However, applications of developed models are usually limited to the conditions they were developed for [21]. As the motion of the human hand has 27 degrees of freedom, a perfect model would not pass the cost-benefits analysis due to amount of calibration required. This would also require recording the signal from multiple muscles. Some grip force estimation models [19, 20] included six muscles (flexors and extensors), however, both of the studies report only minor error reduction if the models included more than three muscles.

Other disadvantages of EMG measurements include complex data processing and the cost of the apparatus. However, ongoing engineering progress permits concurrent attempts [22] to reduce the cost of the hardware. Recent examples of low-cost sEMG applications include rehabilitation exoskeletons [23] and hand prostheses [24, 25]. It is obvious that such solutions will require successful recognition of muscle activation and at least, to some extent, the estimation of strength exertion. The potential of the force exertion estimation still needs assessment.

It is possible to overcome the technological complexity by using modern day low-cost, easy-to-use prototyping platforms such as Arduino [26]. Also, the capability of Arduino microcontrollers to sample EMG signal has been proven [27] and it is possible to purchase commercially manufactured Arduino compatible sEMG measurement shields. A shield is a small electronic board which, if attached to a microcontroller, will add functionality and thus only minimal soldering, if any, is required, which reduces the practitioners' need for an engineering background. Only knowledge about functional anatomy and physiology is mandatory. As the access to sEMG measurements has been improved in the last few years, it is natural that the practitioners, specialized in the domain of physical ergonomics, are interested in the benchmarking of such do-it-yourself (DIY) measurement systems and the task for the scientific community is to provide the assessment.

The aim of this study was to assess the applicability of commercially available low-cost sEMG shields in ergonomics practice. It is hypothesized that Arduino based DIY measurement systems do not perform significantly (α = 0.05) worse than commercially available electromyograph.

2. Methods

2.1. Test objects

In this study, low-cost measurement system is defined as fully operational sEMG measurement apparatus with a total cost below $100. Initially four commercially available sEMG shields were considered – 'Muscle Sensor v3' (Advancer Technologies, Raleigh, USA); 'Olimex Shield EKG/EMG' (OLIMEX Ltd, Plovdiv, Bulgaria), 'Grove EMG Detector' (Seeed Technology Inc, Shenzhen, China) and 'BITalino' (PLUX wireless biosignals, Portugal). All the shields were tested prior to further assessment by recording the sEMG of maximum voluntary contraction from two subjects. Although BITalino showed promising results, it was excluded from further assessment because 1) it is not Arduino compatible; 2) the cost of a fully operational system exceeded the limit of $100. The Grove EMG Detector was also excluded from further assessment because of low signal amplitude and it was impossible to adjust the level of gain.

Finally, Advancer Technologies' Muscle Sensor v3 (adjustable gain, input impedance 0.8 GΩ, CMRR = 90 dB, rectified EMG output) and Olimex Shield EKG/EMG (adjustable gain, input impedance 10 TΩ, CMRR = 80 dB, raw EMG output) were chosen and connected to a 10-bit resolution analog input of an Arduino microcontroller (Smart Projects Srl, Strambino, Italy) for A/D conversion. The performance of these low-cost sEMG measurement systems and the electromyograph ME6000 (Mega Electronics, Kuopio, Finland) was compared in a hand grip task. In order to provide comparable muscle exertion, an electronic dynamometer (Neurosoft, Ivanovo, Russia) was used in this experiment. The dynamometer's I-type handle was an elliptical shape with a circumference of 13.5 cm. Bipolar dual Ag/AgCl disposable electrodes (Noraxon Inc, Scottsdale, USA) were used for sEMG measurements to maintain constant inter-electrode distance (2.0 cm).

2.2. Test subjects and procedure

Ten male subjects with (mean ± SE) age of 24.9 ± 0.7, weight of 80.4 ± 3.3 kg, height of 180.8 ± 1.1 cm and maximum grip force 397 ± 13 N participated in the study. The hand grip force of the dominant hand was measured according to the Caldwell regimen [28] while the forearm was in horizontal position and supported.

Based on initial maximum grip force (isometric contraction), three submaximal exertion levels (25%, 50% and 75% of maximum voluntary contraction) were expected from the participants. The participants were instructed to gradually exert the grip force to the intended level during 1 second and then maintain the effort for another 4 seconds. At least a 2 min break followed each exertion. The electronic dynamometer and its accompanying software provided visual feedback for each submaximal exertion level. The sEMG activity of the m. flexor digitorum superficialis was recorded simultaneously to grip force and normalized to maximum voluntary contraction (MVC). Microsoft Excel was used to calculate normalized sEMG activity (EA) using the following equation [29]:

$$Normalized\ EA = \frac{EMG_{i,j} - MinEMG_j}{MaxEMG_j - MinEMG_j} \tag{1}$$

where $EMG_{i,j}$ is the actual 4 second root mean square (RMS) of EMG signal taken at submaximal exertion level i for subject j

$MinEMG_j$ is the RMS of EMG signal taken at relaxed state for subject j

$MaxEMG_j$ is the 4 second RMS of EMG signal taken from MVC for subject j

As MVC is known to be a motivation dependent variable [30], ANOVA was run in order to compare differences between the MVC-s of three apparatus configurations – the differences were not statistically significant.

3. Results

On average, the normalized EA of the m. flexor digitorum superficialis underestimated exerted grip force (figure 1). Differences between the measurement apparatus were no statistically significant $F(2, 76) = 1.9$, $p = 0.158$.

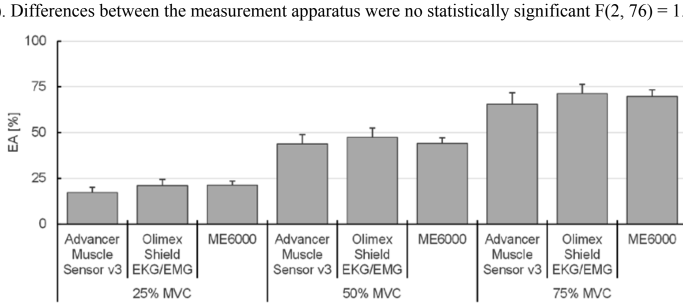

Fig. 1. Electromyographic activity, normalized to MVC, on three submaximal contraction levels (mean + SE).

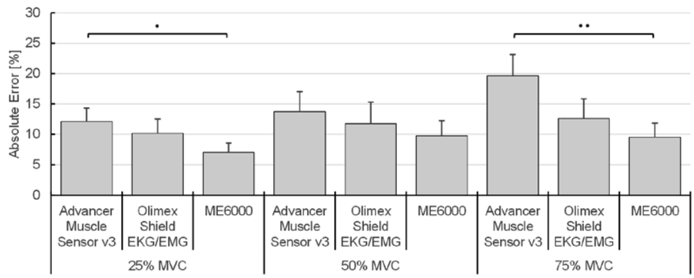

Fig. 2. Absolute differences between normalized EA and relative grip force (mean + SE).

As expected, exertion levels and individual differences of participants had statistically significant effect on EA – $F(2, 76) = 233.0$, $p = 0.000$ and $F(9, 76) = 12.8$, $p = 0.000$ respectively. However, in the case of some of the participants, the normalized EA also overestimated the grip force; therefore, absolute differences between normalized EA and relative grip force were compared (figure 2).

In this case differences in apparatus configurations were found statistically significant $F(2, 76) = 8.8$, $p = 0.000$. Tukey's honest significance test was run within force exertion levels in order to determine the differences. Results (figure 2) showed almost statistically significant ($p = 0.059$) difference between Advancer Technologies' Muscle Sensor v3 and Olimex Shield EKG/EMG at 75% MVC and statistically significant differences between Advancer Technologies' Muscle Sensor v3 and the electromyograph ME6000 at 25% MVC ($p = 0.024$), and 75% MVC ($p = 0.006$). Absolute error was highest at 75% MVC; however, such exertions lies outside the range of most ergonomics studies.

4. Discussion

Both low-cost systems could be used to detect muscle activation and for semi-quantitative assessment of force exertion. There is a certain systematic error in the results as in the case of the gripping task, one muscle is able to explain less variability than multiple muscles and any further investigation should involve at least three muscles as is currently believed to be optimal for a gripping task [19, 20]. However, this error is expected to have equal impact on every configuration of apparatus. Still, the Advancer Technologies' Muscle Sensor v3 was found to be significantly different in the case of absolute error. Potential explanation may be found from the Muscle Sensor v3's product literature. The circuit on the shield contains a high-pass filter with a cutoff frequency of 106 Hz, which is supposed to reduce artifacts from movements or AC current. This could be an advantage in the case of dynamic conditions, but this needs further testing because the current study was focused on isometric contractions. However, as a conventional EMG sensors' range is between 0–400 Hz [31]. Meanwhile the Olimex Shield EKG/EMG is also used for electroencephalography measurements and no effort is made to filter out the lower spectrum of the signal. This might be a weak point in the case of dynamic force exertions which are more prevalent in field conditions.

De Luca [32] has expressed concern that often the correlation between the sEMG amplitude and force exertion might be misleading and if the sEMG is easy to use, it will also be easy to abuse. As the cost restriction to sEMG access is reduced, the sEMG measurements may be more often carried out by so called voodoo-ergonomists. However, the driving force of low-cost sEMG development lies outside the scope of ergonomics, thus the ergonomists' community might either embrace or ignore the potential of low-cost sEMG – it probably won't affect the development of such possibilities. The results of the current study imply that the low-cost sEMG shields tested will allow for the detection of muscle activation (i.e. on-off) patterns and at least for the semi-quantitative assessment of muscle force. Fatigue analysis is restricted as the output of the Advancer Technologies' Muscle Sensor v3is a rectified EMG signal and there is a shortage of easy to apply fatigue assessment routines for raw EMG signal in the case of Olimex Shield EKG/EMG. Although the results are restricted to isometric contractions, the shields tested in this certainly have educational value. Low-cost sEMG shields could be used to provide hands-on experience in order to learn the importance of electrode placement or the need for extensive knowledge about anatomy. Learning to appreciate the virtues and shortcomings of sEMG measurements during different types of contractions could lead to a better understanding of appropriate scientific literature. Thus a low-cost sEMG measurement system might benefit the learning process in life-long learning or on a higher educational level. The Arduino prototyping platform provides the opportunity to measure various factors at a relatively low cost. Simultaneous utilization of sEMG and FSR will provide the opportunity to overcome their shortcomings during force measurements.

Conclusions

This study was conducted in order to assess the potential of low-cost sEMG measurement systems to be applied in ergonomics practice on the example of provided relative grip force estimates. The performances of two low-cost sEMG measurement systems were compared with the performance of a commercially available electromyograph. Both low-cost systems could be used to detect muscle activation and for semi-quantitative assessment of force exertion. The data in this study is limited to isometric contractions and dynamic exertion requires further assessment. Also educational utilization of low-cost sEMG measurement systems is discussed.

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III

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Ergonomic Benefits of an Angle Grinder With Rotatable Main Handle in a Cutting Task.
Human Factors, 61(7), 1112–1124

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Ergonomic Benefits of an Angle Grinder With Rotatable Main Handle in a Cutting Task

Märt Reinvee, **Sander Aia**, Estonian University of Life Sciences, Tartu, Estonia, and **Mati Pääsuke,** University of Tartu, Estonia

Objective: The aim of this study was to evaluate the ergonomic benefits of an angle grinder with a rotatable main handle in a cutting task.

Background: Angle grinder manufacturers rarely address ergonomic features in their advertisements, and if they do, the benefits are expressed in a qualitative manner. Meanwhile, quantitative information about the effects of the device on the worker is required to make informed decisions during tool selection and cumulative trauma prevention.

Method: Eleven maintenance workers and metalworkers used an angle grinder to cut a horizontal steel rod using three wrist postures. Only one of the postures was exclusively available in the case of a rotatable main handle. The postural effect was evaluated objectively with electromyography and a force-sensing-resistor-based force glove. Subjective ratings about discomfort and control were obtained with a visual analog scale.

Results: The subjective ratings favor the near-neutral wrist posture. The forearm muscles' electromyographic activities were similar across the postures. Forces on the hand–handle interface were concentrated on the intermediate phalanges. If the device is operated without gloves, the forces on the intermediate phalanges may exceed the discomfort pressure threshold regardless of wrist posture.

Conclusion: In the cutting task, the subjective measures favor the posture with a near-neutral wrist, which is a feature of the rotatable main handle. The objective measures did not allow one to prefer one posture to another.

Application: The findings give insight into the impact of wrist posture on muscle activity, forces on the hand–handle interface, and discomfort. This is useful information for the person responsible for tool selection.

Keywords: upper extremity, sEMG, tissue loading, tools, physical ergonomics

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INTRODUCTION

In an angle grinder, the rotary motion from the motor drives an abrasive disc, which in metalworking and construction may be used for weld dressing, cutting off a welded part, or slicing bar stock. Although emitting sparks may be the primary visually perceived phenomenon during the abrasive processing of metal, so far the main occupational concerns of angle grinders have been the vibration (Liljelind et al., 2013; Liljelind, Wahlström, Nilsson, Persson, & Nilsson, 2010; Wasserman et al., 2002) and noise exposure. Exposure to vibration increases grip force (Gurram, Rakheja, & Gouw, 1995; Radwin, Armstrong, & Chaffin, 1987), which magnifies the risk of cumulative trauma disorders (CTDs). Moreover, the findings of Adewusi, Rakheja, Marcotte, and Boutin (2010) suggest that grip force and posture have an effect on the vibration exposure; all are known factors of CTDs (Putz-Anderson, 1988).

It is evident that the posture and grip force requirements are to some extent affected by the design of the tool and could be alleviated by ergonomic design. However, a recent survey reported that only a few manufacturers address ergonomic features in their grinding tool advertisements, and the benefits from ergonomic features are mostly described in a nonquantitative manner (Odum, Castillo, Das, & Linke, 2014). Meanwhile, the quantitative information about the ergonomic features would be beneficial, as angle grinders are often used in professions associated with a high (>50%) prevalence of CTDs in the upper extremities (Ahasani, Mohiuddin, Väyrynen, Ironkannas, & Quddus, 1999; Forde, Punnett, & Wegman, 2005).

According to Kroemer (1989), CTDs linked with the use of grinding tools are carpal tunnel syndrome, tendosynovitis, and ganglion. The aforementioned CTDs are associated with radioulnar deviation, repeated wrist flexion or

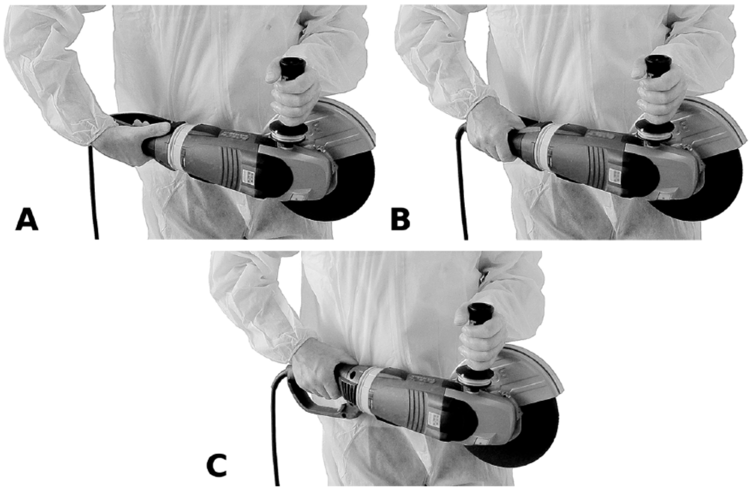

Figure 1. Postures: (A) The bow of the main handle is perpendicular to the abrasive disk; the trigger is activated with four fingers. The posture causes significant wrist flexion. (B) Same as Posture A, but the trigger is activated with the thumb. The posture causes just-noticeable wrist extension. (C) The bow of the main handle is in line with the abrasive disk; the trigger is activated with four fingers. The wrist is in near-neutral position.

extension, and forceful wrist flexion and extension with pressure at the palmar base and pressure on the palm (Kroemer, 1989). Thus the ergonomic assessment of an angle grinder should consider Putz-Anderson's (1988) "faulty tool indicators": excessive or continuous pressure on the palm and fingers, awkward hand position, and static muscle load in the upper arm.

One approach to reduce muscle load and avoid awkward hand positions is to use hand tools that follow the principle, "Bend the tool, not the wrist." Several authors have showed the ergonomic benefits of bent handle tools (Conner & Irwin, 2009; Duke, Mirka, & Sommerich, 2004; Steinhilber et al., 2017), although in some cases, bending the handle does not seem to have any ergonomic benefits (Agostinucci & McLinden, 2016). In this context, the authors investigated the ergonomic benefits of an angle grinder with a 180° rotatable main handle. Usually, grinding is performed with a horizontally positioned disc, and cutting, with a vertically positioned disc. By default, the angle grinders without a rotatable main handle are designed to favor the grinding task, which can be performed in near-neutral posture while the trigger is operated with four fingers. Meanwhile, cutting can be performed in two postures (Figures 1A and 1B).

Both postures have their flaws. Posture A requires extensive wrist flexion, which is known as a risk factor for CTDs. Posture B uses the thumb to activate the trigger, which is less desirable than a four-finger trigger activation (Sanders & McCormick, 1992), and the posture requires slight wrist extension, which may induce a static load in the forearm extensors. A visually more ergonomic posture (Figure 1C) is available in the case of a rotatable main handle. Posture C allows operating the trigger with four fingers while maintaining a near-neutral posture. Postures C and B are used with the coal-hammer grip (Napier, 1956); thus the contributions of individual fingers are expected to be similar. The effect of the Postures A through C on the risk factors for CTDs is not yet known.

This study combines objective and subjective methods to evaluate the effect of the postures on discomfort, control, pressure on the hand, and static muscle load in the forearm muscles. The objective of this study was to determine whether an angle grinder with a rotatable main handle may offer ergonomic benefits in a cutting task.

METHOD

Object and Postures

A professional-grade angle grinder with a rotatable main handle was used in this experiment. The technical characteristics of the device were as follows: weight= 5.5 kg, rated power input= 2.4 kW, no-load speed= 6,500 rpm (≈108 Hz), and disc diameter= 230 mm. The three postures in the experiment (Figure 1) were identified based on the manufacturer's literature and observations in workshops. As the task or posture of the left hand is not affected by Postures A through C, only the effects on the participant's right hand were investigated.

Subjects

The subjects were required to be free from known musculoskeletal disorders, to have hand size corresponding to a size 10 (EN 420) glove, and to have at least 2 years of experience in metalworking occupations. Eleven male participants, with prior experience in metalworking ranging from 2 to 16 years, were recruited from the population of local metalwork companies and self-employed maintenance workers. The participants were engaged in custom metalworking projects and claimed that, depending on the duration and nature of the projects, they used angle grinders on a weekly to monthly basis. None of the participants used angle grinders daily, and the majority of the participants had never used an angle grinder with a rotatable handle.

The means of the participants' anthropometric measurements and muscle strength were as follows: age, 30.5 (SE = 2.1) years; height, 181.0 (SE = 2.2) cm; weight, 84.1 (SE = 2.8) kgl; power grip strength, 421 (SE = 73) N; and lateral pinch grip strength, 147 (SE = 22) N. Each participant provided their informed consent after being briefed about the goals and procedures of the experiment. In the case where the tests were conducted on a company worksite, written consent was obtained from the management.

Instrumentation and Apparatus

Dynamometry. An I-type dynamometer (Vernier Software & Technology, USA), with a circumference of 155 mm, was connected to a 10-bit Arduino Uno microcontroller (Arduino, Italy). An in-house developed visual feedback system (three preprogrammed and remote-controlled LEDs) was used to assist the participants in following the main test procedure and Caldwell et al.'s (1974) regimen during dynamometry.

Electromyography (EMG). A four-channel 10-bit BITalino telemetric microcontroller (PLUX Wireless Biosignals, Portugal) was used in this experiment. Three channels were equipped with EMG 151015 amplifiers (PLUX Wireless Biosignals, Portugal). The bipolar differential amplifiers had gain of 1000 V/V, input impedance of 100 GΩ, common-mode rejection ratio of 100 dB, and bandwidth of 10 to 400 Hz. Dual Ag/AgCl disposable electrodes (Noraxon, USA), with a constant interelectrode distance of 2.0 cm, were placed on the muscle bellies of the flexor carpi ulnaris (FCU), flexor carpi radialis (FCR), and extensor digitorum (ED). The skin under the electrodes was shaved and cleaned with alcohol.

The muscles where chosen based on the approach of Kadefors et al. (1993); the sites on the muscles were chosen in accordance to Cram, Kasman, and Holtz (2011); and the locations were confirmed manually by palpation and visually by inspecting the data corresponding to the wrist extension and flexion in real time. The remaining fourth channel of the microcontroller was equipped with a single-axis accelerometer. The accelerometer's data set was used for the solemn purpose of distinguishing loaded and unloaded periods in the EMG data set.

The EMG and accelerometer data were acquired at a sample rate of 1000 Hz with OpenSignals (r)evolution software (PLUX Wireless Biosignals, Portugal). The raw EMG data were processed with a custom MATLAB (MathWorks, USA) script. The script removed the DC-offset and used a Butterworth second-order low-pass filter, with cutoff frequency set to 2 Hz in order to calculate the linear envelope.

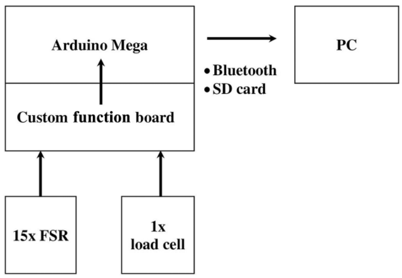

Figure 2. Block diagram of the force glove system.

Force glove system. A force glove system was constructed based on the example of Lowe, Kong, and Han (2006). The force glove system was based on open-source hardware and software (Figure 2). The main element of the system was a 16-channel, 10-bit Arduino Mega ADK microcontroller (Arduino, Italy), which was used for programmable analog-digital conversion. A custom-built function board connected to the microcontroller allowed storing or transmitting the data and changing the sensitivity and range of the sensors. The number of sensors used in the force glove was limited by the number of channels on the microcontroller. One channel was used to connect a button-type load cell model C2 (HBM, Germany) in order to calibrate the force-sensing resistors (FSRs). The FSRs (model FlexiForce A201, Tekscan, USA) were attached to a size 10 leather glove with 3M Micropore tape. The FSR locations (Figure 3) were chosen based on the analysis of Reinvee and Jansen (2014), and the sensors were calibrated before each testing session similarly to the procedure described by Kong and Lowe (2005).

First, the load cell was calibrated using dead weights, and the output of the load cell was linearly regressed to the weights (R^2 = 1.000 in all cases). The FSRs were placed one by one between the investigator's thumb and the load cell; the force was then gradually increased from 0 to 25 N (a limit chosen based on preliminary testing). This procedure was repeated thrice, and third-order polynomial regressions were used to determine the relationship ($.974 \leq R^2 \leq .998$) between the force and the FSR output.

The force glove system operated with a sample rate of 10 Hz, and the data were processed in MATLAB. The processed data were turned into cumulative distribution functions of force (CDF_Fs), an approach proposed by Yun, Kotani, and Ellis (1992). The x in the $CDF_F(x)$ is the probability that a force value is less than or equal to a certain value. According to Jonsson (1982), the $CDF_F(.1)$ corresponds to static, $CDF_F(.5)$ to dynamic, and $CDF_F(.9)$ to peak load.

Subjective ratings. Subjective ratings were obtained with 100-mm visual analog scales (VAS). Each VAS consists of a straight line with verbal anchors on both ends of the scale. No other visual cues (numbers or tick marks) besides the verbal anchors were provided for the participants. The distance in millimeters from the left anchor to the participant's mark on the scale was counted as the rating. Only the ratings of control and discomfort were collected in this study. These ratings were considered to suffice, as the experimental procedure was intended to be short, robust, and minimally fatiguing in order to avoid hampering the work of other workers in the workshops.

Experimental Procedure

The procedure complied with the tenets of the Declaration of Helsinki and was approved by the local ethics committee. After being briefed and providing their informed consent, the participants were guided to practice the task. The practice session was followed by a 5- to 10-min break during which the experimenters attached the EMG electrodes and secured the cables. The EMG signal was normalized to maximum voluntary contraction (MVC) during self-restricted wrist flexion and extension. The principles of Caldwell's regimen (Caldwell et al., 1974) were followed both to obtain the MVCs and the grip strengths. The participants were trained to increase the exertion smoothly during a period of 1 s and then sustain a steady exertion for 4 s. The averages from the 2-s period in the middle of the steady exertion were used as the grip strength and MVC values. In the case of MVCs, the higher value, and in the case of grip strengths, the average value of two attempts, was used.

After obtaining the MVCs, the participants were fitted with the force glove system and started on a three-task work sequence. In the three-task work sequence (Figure 4), the

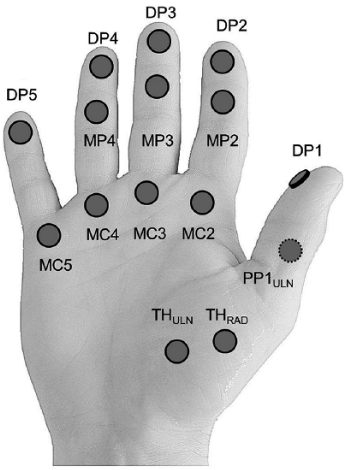

Figure 3. Force-sensing resistors' locations on the hand. DP = distal phalanges; MP = intermedial phalanges; PP = proximal phalanges; MC = metacarpals; TH = thenar region. Numbers 1 to 5 represent the fingers from thumb to little finger; subscripts indicate the ulnar or radial location of the sensor.

participants had to (1) pick up and hold the device, (2) start the device and let the motor run on full speed, and (3) cut three pieces from a horizontal Ø10-mm steel rod with Rockwell hardness of 77.2 HRB (SE = .4). The steel rod was clamped between the jaws of metalworking vices, and the available length from the attachment was shortened from 10 to about 5 cm from the first to the last cut. The height of the attachment varied from 102 to 108 cm across the test sites. The duration of the sequence was about 1 min, and the pace of the first two tasks was indicated and kept constant with an experimenter-controlled visual feedback system. The sequence was then repeated in the next posture after a 2-min break, and the order of postures was randomized between the participants.

After participants completed the three-task work sequence in all three postures, the subjective ratings were obtained. The objective measures were obtained in accordance to the scheme in Figure 4. The accelerometer was attached to the main handle, and the data set was band-pass filtered to distinguish no-load and loaded periods. Each EMG sample was the average of a 2-s period. The force glove system's sample started 2 s before pressing the trigger and ended 1 s after finishing the third cut.

Data Analysis

The Shapiro-Wilk and Fligner-Killeen tests were used to test the normality and homoscedasticity of the data. As ANOVA assumptions were not met, the following nonparametric tests were used: the Friedman test for one-way ANOVA and the Wilcoxon rank-sum test with the Benjamini and Hochberg procedure for pairwise comparisons. Statistical analysis was conducted with R version 3.4.3 (R Development Core Team). The level of statistical significance was set to $\alpha = .05$. Data are reported as means and standard errors of the mean.

The analysis of EMG data was conducted with $n = 10$, as the data from one person were corrupt. Also, in one case, an FSR attached to the intermedial phalange of the middle finger (MP3) broke down, and the data from one posture was lost. As the Friedman test was not applicable in this case due to an incomplete data set, only post hoc test was used.

RESULTS

Subjective Ratings

The participants' subjective ratings (Figure 5) were statistically significantly different both in the case of discomfort, $\chi^2(2) = 15.954$, $p < .001$, and control, $\chi^2(2) = 17.636$, $p < .001$. However, the pairwise comparison revealed that Posture C differed in both cases from Postures A and B ($p = .001$), but the difference between Postures A and B was not statistically significant either in the case of discomfort ($p = .056$) or in the case of control ($p = .308$).

EMG

Normalized electromyographic (nEMG) activities were compared within the muscle and between postures (Figure 6). In the case of FCU and FCR, the posture did not have a statistically significant effect on nEMG in any of the tasks studied. In the case of ED, nEMG was

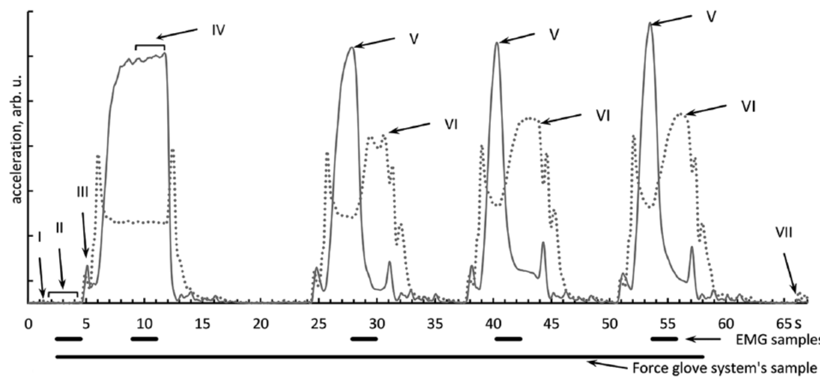

Figure 4. Data processing scheme. Solid gray line = acceleration data, band-pass filtered in range 100 to 110 Hz; dotted gray line = acceleration data, band-pass filtered in range 60 to 95 Hz; I = picking up the angle grinder; II = static holding of the device; III = pressing the trigger; IV = full speed, no load; V = start of the cut; VI = releasing the trigger; VII = putting the device away.

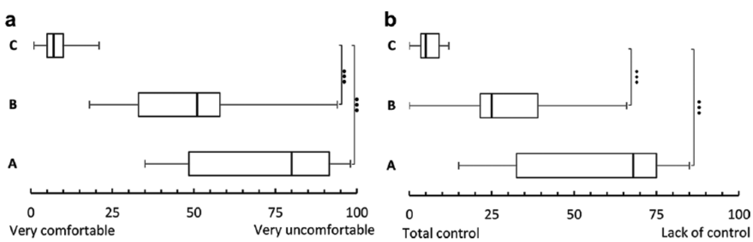

Figure 5. Subjective ratings of (a) discomfort and (b) control of the Postures A through C (see Figure 1). The box spans the interquartile range, the line inside the box indicates the median rating, and error bars indicate the lowest and highest ratings. ****p* < .001.

statistically significantly affected by posture, $\chi^2(2) = 15.000$, $p = .001$, while holding the angle grinder.

Force Glove System

An uneven distribution of interface pressure among hand regions is evident from the CDF_Fs in Figure 7. Meanwhile, only a few locations showed a statistically significant effect of posture on force on either static, dynamic, or peak load level (Table 1). On static load level, $CDF_F(.1)$, only one location (MC5) was found with a statistically significant postural effect; however, the adjusted probabilities in pairwise comparison were found not significant ($p \geq .36$). Figure 8 shows the results of pairwise comparisons on dynamic and peak load levels.

DISCUSSION

In this study we investigated the ergonomic benefits of an angle grinder with a rotatable main handle and focused on the objective and subjective effects of three postures during a cutting task. The subjective measures (Figure 5) indicate a preference toward Posture C, which is a feature of the rotatable main handle. This is not

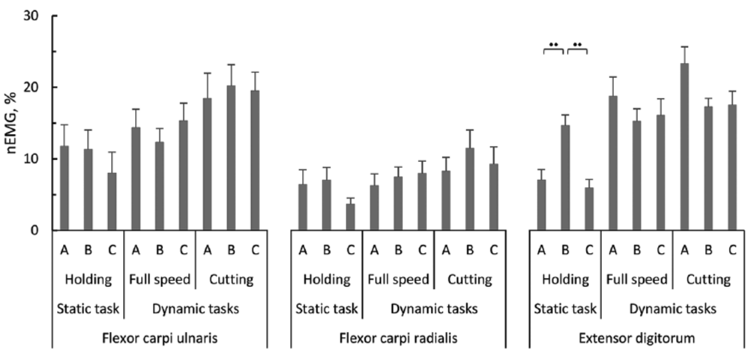

Figure 6. Normalized electromyographic activities ($M + SE$) according to posture and task. See Figure 1 for Postures A through C. $^{**}p < .01$.

surprising as deviations from the neutral posture are known to reduce grip strength capacity (Terrell & Purswell, 1976); thus the operator may sense an increase in physical demands, which is related to control and discomfort. Posture B seems visually acceptable (Figure 1), yet it requires the thumb to operate the trigger, which is deemed less desirable than using four fingers (Sanders & McCormick, 1992). Furthermore, the thumb's strength capacity allows it to contribute less than half (<38%) of the total grip force (Kinoshita, Murase, & Bandou, 1996).

There is a potential bias in the subjective ratings, as the ratings were collected at the end of the experiment. We believe that the potential bias does not discredit the superiority of Posture C in the participants' ratings, as the participants practiced the task prior to the experiment in all postures and the order of postures was randomized. In this context, the insignificant difference in the ratings of control ($p = .308$) between Postures A and B is not surprising. Meanwhile, the differences in discomfort ratings ($p = .056$) may or may not exceed the level of significance in a larger test group or in the absence of the potential bias.

The skepticism is based on the EMG of the ED muscle (Figure 6), as forearm extensors are known to be more prone to fatigue than flexors (Hägg & Milerad, 1997). One of the main functions of ED is to extend the wrist, which is evident in Posture B. During the static task in Posture B, the ED shows a significantly larger muscle load than during the same tasks in Posture A or C ($p = .002$ in both cases). Meanwhile, the two muscles on the volar side of the forearm do not show a significant postural effect in the case of the flexed-wrist Posture A. Moreover, pressing the trigger requires thumb adduction; thus the picture is incomplete without the EMG of the adductor pollicis (AD) muscle.

The absence of the EMG of the AD or its antagonistic muscles is a shortcoming of this study, which relates to the experimental design and available instrumentation. First, the AD location was not directly accessible (due to discomfort for the participants and signal-contaminating artifacts); also the subjects wore gloves, which restricted the measurement of other agonistic or antagonistic muscles. Second, the EMG acquisition device had four channels. One of the channels was occupied by a single-axis accelerometer that allowed one to measure only three muscles. Angle grinders are rarely the only tool used by metalworking professionals. Thus, prioritizing the forearm muscles, as proposed by Kadefors et al. (1993), seemed to serve a holistic view of forearm muscle load.

Still, the necessary insight about thumb loading was obtained from the FSR attached to the ulnar side of the thumb's proximal phalange ($PP1_{ULN}$). On average, the CDF_Fs on $PP1_{ULN}$ (Figure 7) were rather similar in the case of

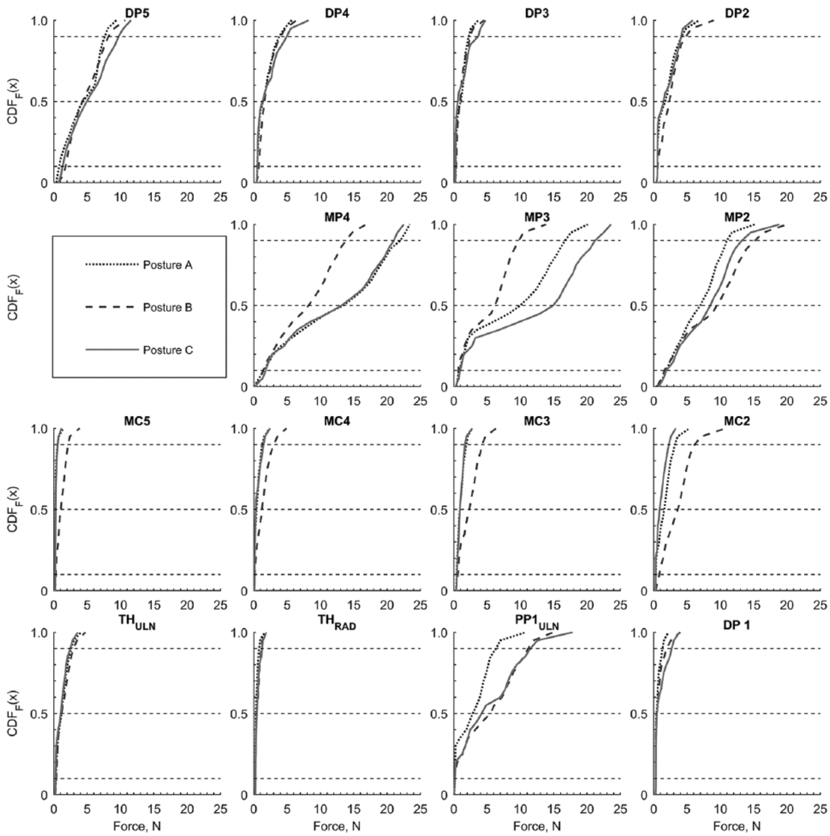

Figure 7. Averaged cumulative distribution functions of force. For Postures A through C, see Figure 1; for sensor locations, see Figure 3.

Postures B and C; furthermore, both showed higher force exertions than in Posture A. Although this finding potentially indicates the importance of thumb adduction in controlling the device, the current study does not provide evidence for the case, as the differences between postures were not statistically significant (Table 1). As a significantly higher thumb force exertion was expected in the case of Posture B than in the cases of Postures A and C, future studies should either include the EMG of thumb muscles or increase the number of FSRs attached to the thumb.

Meanwhile, statistically significant postural effects were found from the forces on intermediate phalanges and metacarpals (Table 1). Nevertheless, in a practical sense, the differences on metacarpals are trivial. This is a passive region; the forces are low (Figures 7 and 8) and will not cause inconvenient pressure points. The increase in force in the case of Posture B is caused by the region's opposition to the thumb in this posture. In contrast, the intermediate phalanges are active regions with major force contribution. Also, there is a noticeable increase in force exertion from static load level to dynamic load level

TABLE 1: Results of the Friedman Test Indicating Differences in Force on Hand Regions and Load Levels (Static, Dynamic, and Peak) Between Variants of Posture

	$CDF_F(.1)$		$CDF_F(.5)$		$CDF_F(.9)$	
FSR Location	$\chi^2(2)$	p	$\chi^2(2)$	p	$\chi^2(2)$	p
DP5	0.884	.643	3.116	.211	3.767	.152
DP4	4.546	.103	2.651	.266	2.637	.307
DP3	2.889	.236	2.000	.368	1.273	.529
DP2	0.546	.761	3.818	.148	1.273	.529
MP4	1.400	.497	**6.727**	**.035**	**11.302**	**.004**
MP3	n/a	n/a	n/a	n/a	n/a	n/a
MP2	0.182	.913	2.909	.234	**7.091**	**.029**
MC5	**7.659**	**.022**	**10.947**	**.004**	**13.905**	**.001**
MC4	0.400	.819	**7.818**	**.020**	**6.727**	**.035**
MC3	2.364	.307	**8.909**	**.012**	**8.727**	**.013**
MC2	3.455	.178	**15.273**	**<.001**	**15.273**	**<.001**
TH_{ULN}	4.546	.103	2.364	.307	2.182	.336
TH_{RAD}	2.889	.236	1.857	.395	1.273	.529
$PP1_{ULN}$	0.250	.883	2.279	.320	1.333	.514
DP1	2.000	.368	0.438	.804	**6.727**	**.035**

Note. Values in bold indicate a statistically significant effect of posture; *n/a* indicates a force-sensing resistor location with an incomplete data set. FSR = force-sensing resistor; CDF_F = cumulative distribution function of force; DP = distal phalange; MP = intermedial phalange; MC = metacarpal; TH = thenar region; PP = proximal phalange. Numbers 1 to 5 represent the fingers from thumb to little finger; subscripts indicate the ulnar (ULN) or radial (RAD) location of the sensor.

(Figure 7), which is in line with the findings of Gurram et al. (1995). These values were compared with the submaximal finger forces in an experiment of Kong, Lee, Kim, and Jung (2011) and were found to be in range of 10% to 20% MVC. In the experiment of Kong et al., the individual finger contributions were obtained with a straight wrist; thus only Posture C is directly comparable. However, this comparison indicates an agreement between the results from EMG (Figure 6) and the force glove system.

In addition, using the FSR diameter (9.7 mm), a force equivalent to the finger's discomfort pressure threshold (Johansson, Kjellberg, Kilbom, & Hagg, 1999) was calculated and compared with the CDF_Fs of the intermedial phalanges. A considerable proportion of the CDF curves corresponding to Postures A and C (Figure 7) exceeded the calculated threshold criteria, 13.9 N. However, there are two issues when drawing conclusions related to the objective criteria of discomfort. On one hand, the discomfort pressure threshold values in Johansson et al. (1999) were obtained from bare-handed white-collar workers, and in the current study, the participants were glove-handed blue-collar workers. On the other hand, the participants were instructed not to exert unnecessary force during the experiment; multiple participants claimed that in the case of high temporal demands, they would exceed their current effort. All in all, based on these considerations, it is not necessary to distinguish the postures in regard to discomfort pressure threshold; however, this may be an issue in the presence of a CTD, which significantly reduces the threshold (Gold, Punnett, & Katz, 2006).

The second issue related to the application of the results is related to the postures. As the study was conducted on several sites in the field, the on-site inventory dictated the fixed heights of the steel rods, which caused variations in elbow and wrist posture due to differences in the participants' heights. Meanwhile, the postures were

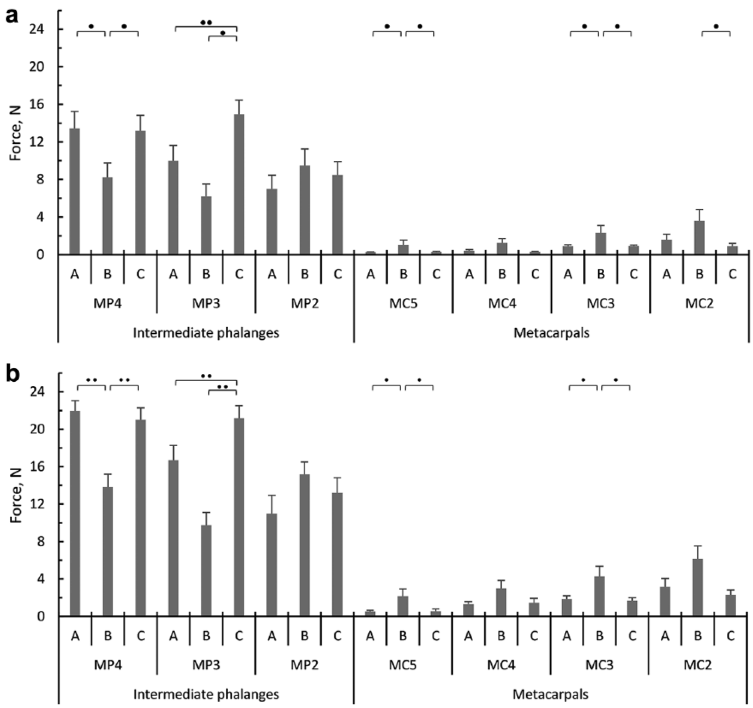

Figure 8. Forces on force-sensing resistors ($M + SE$) attached to the intermediate phalanges and metacarpals: (a) dynamic load levels, $CDF_F(.5)$; (b) peak load levels, $CDF_F(.9)$. See Figure 1 for Postures A through C; for sensor locations, see Figure 3. CDF_F = cumulative distribution function of force. $*p < .05$. $**p < .01$.

not measured, which did not allow one to assess the effect of postural differences within the postures, nor did it allow one to assess the results in a wider context, such as comparing the results with similar devices (e.g., some operations with chainsaws). Yet, the three postures (Figure 1) are clearly distinguishable without the exact angular values. Also, during the experiment, one participant pointed out that in the absence of the rotatable handle, sometimes a third posture is used instead of Posture A or B. In the case of the "alternative" posture, the purpose of the left and right hand is reversed (Figure 9).

A safety specialist was consulted about the rationale of the alternative posture. In the case of high temporal demands, the grinding and cutting tasks are sometimes performed without changing the task-specific guard. The type 27 guard for the grinding task covers half of the abrasive disk from one side; meanwhile, the type 1 guard for the cutting task covers half from both sides of the abrasive disk. If the trajectory of sparks is not restricted, then the alternative posture serves as a safety measure. Also, there are claims that the preference toward the alternative posture is increased when workers have first- or secondhand experience of abrasive disk breakage or they suffer from a CTD in their right hand (M. Luik, personal communication, February 20, 2018). The participants in this study were right-handed and claimed to be free from CTDs; however,

Figure 9. Front and top view of the "alternative" posture. Similarly to Postures A and B (Figure 1), the bow of the main handle is perpendicular to the abrasive disk. The type 27 guard on the angle grinder is for the grinding task.

no assumptions can be made about their first-hand experience of abrasive disk breakage. The participants complied with the procedure and did not give any noteworthy comments about Posture B, whereas Posture A was in some cases referred to as the posture of a "rookie" or an "apprentice."

Further investigation is needed to analyze the potential ergonomic benefits of the alternative posture. This posture cannot be recommended just on the basis of questionable safety improvements during the usage of improper guards. However, some cutting tasks in construction might benefit from the alternative posture with proper cutting guards (M. Luik, personal communication, February 20, 2018). However, reversing the hands' purpose in the current cutting task might have an effect on proprioception. The postural changes may affect the visual conditions, which can hinder the ability to aim with the tool (Chintapalli, Ding, & Hallbeck, 2003). Thus, the investigation of the alternative posture should include the measuring of performance in addition to the objective and subjective effects on the operator. Finally, further studies should also consider the effect of hand size. The hand size in this study was a controlled variable and limited to EN 420 size 10; thus extrapolation of the values from the subjective or objective measures to other hand sizes should be made with caution.

CONCLUSION

The near-neutral wrist posture in a cutting task is a feature of the rotatable main handle. The results of the subjective ratings favor the near-neutral wrist posture. The muscular load, measured with EMG, and the forces measured on the hand–handle interface did not allow one to prefer one posture to another. Further investigation is needed to determine the most ergonomic posture in the absence of the rotatable handle.

ACKNOWLEDGMENTS

The authors are grateful to safety specialist Matis Luik for sharing his experience from fieldwork.

KEY POINTS

- Posture C, which was exclusively available in the case of a rotatable main handle, was unanimously perceived as superior to Postures A and B. Postures A and B received diverse ratings.
- During the cutting task, the electromyographic activity of the flexcor carpi radialis muscle (~10% maximum voluntary contraction [MVC]) did not differ significantly in the case of the three postures, nor did the electromyographic activities of the flexcor carpi ulnaris and extensor digitorum muscles (~20% MVC).
- Forces measured on the hand–handle interface were concentrated on the intermediate phalanges.

The exerted forces may potentially exceed the discomfort pressure threshold when the angle grinder is operated without gloves.

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Märt Reinvee is a PhD candidate at the Estonian University of Life Sciences in the Engineering Sciences Program. He received an MSc in ergonomics from Estonian University of Life Sciences in 2009.

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CURRICULUM VITAE

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Education

2012-2020	PhD studies, Estonian University of Life Sciences
2009-2011	Vocational education in Electricity, Tartu Vocational Education Center
2007-2009	Master's degree in Ergonomics, Estonian University of Life Sciences (*cum laude*).
2004-2007	Bachelor's degree in Engineering, Estonian University of Life Sciences
1992-2004	Räpina Co-Educational Gymnasium

Assigned profession

2011	Electrician 1st level certificate No. 061336

Career

2011…	Estonian University of Life Sciences, Institute of Technology, Specialist

Software skills

Planning	AutoCad, SolidWorks, Dialux
Modelling	MatLab, NoiseAtWork
Graphics	Gimp
Statistics	R
Office	MS Office

Languages

Estonian	Native speaker
English	Excellent in spoken and written language
German	Good in spoken language
Russian	Basic skills
Polish	Basic skills

Participation in professional societies

2009-…	Estonian Ergonomics Society
2014-…	Human Factors and Ergonomics Society

Teaching

2011-2012	TE.0353 Working environment in industry and in agriculture (only practical sessions)
2012	TE.0006 Fundamentals of technology design
2012-…	TE.0874 Occupational safety and ergonomics
2014-2018	TE.0533 Noise and vibration (only practical sessions)
2016-2017	TE.0925 Ergonomics methods
2018-…	TE.0967 Ergonomic assessment of work environment
2018-…	TE.0971 Design of work environment
2018-…	TE.0968 Physical ergonomics

Supervised theses

2019	Indrek Avi, (sup) **Märt Reinvee**, Digital learning material for the prevention of musculoskeletal disorders, Master's Thesis.
2019	Mari-Liis Štrik-Ott, (sup) **Märt Reinvee**, Development of electromyography biofeedback device, Master's Thesis.
2018	Sander Aia, (sup) **Märt Reinvee**, Normalisation of muscle's bioelectrical activity in assessment of ergonomic quality of hand tool, Master's Thesis.

2017	Helari Kukk (sup) **Märt Reinvee**, Hand grip force measurement system based on Arduino microcontroller, Graduation thesis of professional higher education
2016	Vahur Luik, (sup) **Märt Reinvee**, Test rig for the assessment of pinch grip's strength parameters, Master's Thesis.
2015	Peeter Vaas, (sup) **Märt Reinvee**, Development of affordable electromyograph, Master's Thesis.
2015	Indrek Lepasaar, (sup) **Märt Reinvee**; Ada Traumann, The effect of gloves on manual dexterity, Master's Thesis.
2015	Liana Lodi, (sup) **Märt Reinvee**, Ergonomic criteria for the evaluation of hand tools, Bachelor's Thesis.
2014	Rauno Tamm, (sup) **Märt Reinvee**, Anthropomorphic test rig of noise and noise protection devices, Master's Thesis.
2013	Signe Oru, (sup) **Märt Reinvee**, Conformity of hand held tools, Master's Thesis.
2013	Riin Raimla, (sup) **Märt Reinvee**, Ergonomics of working environment in the beauty service company, Bachelor's Thesis.
2013	Svetlana Mohhareva, (sup) **Viljo Viljasoo**; Märt Reinvee, Sitting Postures Ergonomics, Bachelor's Thesis.
2012	Regina Lensment, (sup) **Märt Reinvee**; Arvo Leola, Ergonomicness of pipeline milking machines, Master's Thesis.
2012	Sander Aia, (sup) **Märt Reinvee**, Working environment in company Eben Puit, Bachelor's Thesis.
2012	Lauri Kivimaa (sup) **Märt Reinvee**, Work environment at the woodworking department of company Ilmre, Bachelor's Thesis.
2012	Rauno Tamm, (sup) **Märt Reinvee**, Ergonomics of tablet personal computer and e-reader, Bachelor's Thesis.

ELULOOKIRJELDUS

Kontaktandmed

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Sünniaeg:	27.05.1985
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Haridustee

2012-2020	Doktoriõpe, Eesti Maaülikool
2009-2011	Kutseõpe elektriku õppekaval, Tartu Kutsehariduskeskus
2007-2009	Magistriõpe ergonoomika õppekaval (*cum laude),* Eesti Maaülikool
2004-2007	Bakalaureuseõpe, tehnika ja tehnoloogia õppekaval, Eesti Maaülikool
1992-2004	Räpina Ühisgümnaasium

Omistatud kutsed

2011	Elektrik I, kutsetunnistus nr. 061336

Töökogemus

2011…	Eesti Maaülikooli tehnikainstituut, spetsialist

Arvutioskused

Projekteerimine	AutoCad, SolidWorks, Dialux
Modelleerimine	MatLab, NoiseAtWork
Graafika	Gimp
Statistika	R
Kontoritarkvara	MS Office

Keeled

Eesti keel	Emakeel
Inglise keel	Väga hea nii kõnes kui kirjas
Saksa keel	Hea kõnes
Vene keel	Algtase
Poola keel	Algtase

Organisatsiooniline tegevus

2009-…	MTÜ ErgoEst
2014-…	Human Factors and Ergonomics Society

Õppetöö

2011-2012	TE.0353 Töökeskkond tööstuses ja põllumajanduses (praktikumid)
2012	TE.0006 Tehnoloogia projekteerimise alused
2012-…	TE.0874 Kutseohutus ja ergonoomika
2014-2018	TE.0533 Müra ja vibratsioon (praktikumid)
2016-2017	TE.0925 Uurimismeetodid ergonoomikas
2018-…	TE.0967 Töökeskkonna ergonoomikaline hindamine
2018-…	TE.0971 Töökeskkonna projekteerimine
2018-…	TE.0968 Füüsiline ergonoomika

Juhendatud lõputööd

2019	Indrek Avi, (juh) **Märt Reinvee**, Digitaalne õppematerjal luu- ja lihaskonna ülekoormushaiguste ennetamiseks, magistritöö.
2019	Mari-Liis Štrik-Ott, (juh) **Märt Reinvee**, Elektromüograafia tagasisideseadme arendus, magistritöö.
2018	Sander Aia, (juh) **Märt Reinvee**, Lihaste bioelektrilise aktiivsuse normaliseerimine tööriista ergonoomilisuse hindamisel, magistritöö.

2017	Helari Kukk (juh) **Märt Reinvee**, Käe pigistusjõu mõõtmissüsteemi arendus Arduino mikrokontrolleri baasil, rakenduskõrgharidusõppe lõputöö
2016	Vahur Luik, (juh) **Märt Reinvee**, Täppishaarde jõuparameetrite uurimisstend, magistritöö.
2015	Peeter Vaas, (juh) **Märt Reinvee**, Kuluefektiivse elektromüograafi arendus, magistritöö.
2015	Indrek Lepasaar, (juh) **Märt Reinvee**; Ada Traumann, Kinnaste mõju käelisele osavusele, magistritöö.
2015	Liana Lodi, (juh) **Märt Reinvee**, Käsitööriistade ergonoomilisuse hindamiskriteeriumid, bakalaureusetöö.
2014	Rauno Tamm, (juh) **Märt Reinvee**, Antropomorfne müra ja mürakaitsevahendite uurimisstend, magistritöö.
2013	Signe Oru, (juh) **Märt Reinvee**, Käsitööriistade konformsus, magistritöö.
2013	Riin Raimla, (juh) **Märt Reinvee**, Töökeskkonna ergonoomika iluteenindusettevõttes, bakalaureusetöö.
2013	Svetlana Mohhareva, (juh) Viljo Viljasoo; **Märt Reinvee**, Istumisasendite ergonoomika, bakalaureusetöö.
2012	Regina Lensment, (juh) **Märt Reinvee**; Arvo Leola, Torusselüpsimasinate ergonoomilisus magistritöö.
2012	Sander Aia, (juh) **Märt Reinvee**, Töökeskkond ettevõttes Eben Puit, bakalaureusetöö.
2012	Lauri Kivimaa (juh) **Märt Reinvee**, Töökeskkond ettevõtte Ilmre puidutöötlusosakonnas, bakalaureusetöö.
2012	Rauno Tamm, (juh) **Märt Reinvee**, Tahvelarvuti ja e-lugeri ergonoomika, bakalaureusetöö.

LIST OF PUBLICATIONS

Classification	1		2	3	
Number of publications	10		1	6	
Sub-classification	1.1.	1.2.	2.5.	3.1.	3.4.
Number of publications	8	2	1	5	1

1.1. Scholarly articles indexed by Web of Science Science Citation Index Expanded, Social Sciences Citation Index, Arts & Humanities Citation Index and/or indexed by Scopus (excluding chapters in books)

Luha, A., Merisalu, E., **Reinvee**, **M.**, Kinnas, S., Jõgeva, R., & Orru, H. (*in press*). In-vehicle noise exposure among military personnel depending on type of vehicle, riding compartment and road surface. Journal of the Royal Army Medical Corps.

Reinvee, **M.**, Aia, S., & Pääsuke, M. (2019). Ergonomic Benefits of an Angle Grinder With Rotatable Main Handle in a Cutting Task. Human Factors: The Journal of the Human Factors and Ergonomics Society, 61(7), 1112–1124.

Bergmann, M., Zahharova, A., **Reinvee**, **M.**, Asser, T., Gapeyeva, H., & Vahtrik, D. (2019). The Effect of Functional Electrical Stimulation and Therapeutic Exercises on Trunk Muscle Tone and Dynamic Sitting Balance in Persons with Chronic Spinal Cord Injury: A Crossover Trial. Medicina, 55(10), 619.

Reinvee, **M.**, & Jansen, K. (2014). Utilisation of tactile sensors in ergonomic assessment of hand-handle interface: a review. Agronomy Research, 12(3), 907–914.

Jansen, K., Luik, M., **Reinvee**, **M.**, Viljasoo, V., Ereline, J., Gapeyeva, H., & Pääsuke, M. (2013). Hand discomfort in production assembly workers. Agronomy Research, 11(2), 407–412.

Reinvee, **M.**, Uiga, J., Tärgla, T., Pikk, P., & Annuk, A. (2013). Exploring the effect of carbon dioxide demand controlled ventilation system on air humidity. Agronomy Research, 11(2), 463–470.

Reinvee, M., Luik, M., & Kliimak, P. (2013). Noise emission from grain dryers and potential noise pollution. Agronomy Research, 11 (2), 457−462.

Kliimak, P., Luik, M., Leola, A., & **Reinvee, M**. (2012). Analysis of students' assessments about the noise educational lecture. Agronomy Research, 10 (1), 131−138.

1.2. Peer-reviewed articles in other international research journals with an ISSN code and international editorial board, which are circulated internationally and open to international contributions

Pomerants, T., Viljasoo, V., & **Reinvee, M**. (2013). Maasoojuspump Booster SP talitlus-tehnilised karakteristikud ja ruumi sisekliima. Agraarteadus, XXIV (1), 21−28.

Jansen, K., Luik, M., **Reinvee, M**., Viljasoo, V., Ereline, J., Gapeyeva, H., & Pääsuke, M. (2012). Musculoskeletal discomfort in production assembly workers. Acta kinesiologiae Universitatis Tartuensis, 18, 102−110.

2.5. Published research project report or study

Sirge, T., Merisalu, E., Raimla, R., **Reinvee**, **M**., & Teras, E. (2017). Ruumi tehisvalgustuse mõju tööviljakusele. Tartu.

3.1. Articles/chapters in books published by the publishers listed in Annex (including collections indexed by the Thomson Reuters Book Citation Index, Thomson Reuters Conference Proceedings Citation Index, Scopus)

Reinvee, M., & Mrugalska, B. (2019). Contemporary Low-Cost Hardware for Ergonomic Evaluation: Needs, Applications and Limitations. In W. Arwowski, S. Trzcielinski, B. Mrugalska, M. Di Nicolantonio, & E. Rossi (Eds.), Advances in Manufacturing, Production Management and

Process Control. AHFE 2018. Advances in Intelligent Systems and Computing (Vol. 793, pp. 386–397). Cham: Springer.

Aia, S., & **Reinvee**, **M**. (2018). An investigation of forearm EMG normalization procedures in a field study. In P. M. Arezes, J. S. Baptista, M. P. Barroso, P. Carneiro, P. Cordeiro, N. Costa, … G. Perestrel (Eds.), Occupational Safety and Hygiene VI (pp. 313–317). Guimarães: CRC Press/Balkema.

Reinvee, **M**., & Pääsuke, M. (2016). Overview of Contemporary Low-cost sEMG Hardware for Applications in Human Factors and Ergonomics. Proceedings of the Human Factors and Ergonomics Society Annual Meeting, 60(1), 408–412.

Reinvee, **M**., Vaas, P., Ereline, J., & Pääsuke, M. (2015). Applicability of Affordable sEMG in Ergonomics Practice. Procedia Manufacturing, 3, 4260–4265.

Leola, A., **Reinvee**, **M**., Luik, M., Lensment, R., & Leola, T. (2013). Assessing ergonomicness of pipeline milking machines - methodological approach. In L. Malinovska & V. Osadcuks (Eds.), Proceedings of the 12th International Scientific Conference Engineering for Rural Development (pp. 132–136). Jelgava: Latvia University of Agriculture.

3.4. Articles/presentations published in conference proceedings not listed in Section 3.1

Sirge, T., Teras, E., **Reinvee**, M., Merisalu, E., Raimla, R., Ereline, J., Gapeyeva, H., Kums, T., & Pääsuke, M (2017). Kontoritöötajate skeleti-lihassüsteemi vaevused ja töövõime. *Eesti Arst, 96(Lisa 1), 1–64.*

CONFERENCE PRESENTATIONS

12th International Scientific Conference Engineering for Rural Development, 23.-24.05.2013, Jelgava, Latvia, Assessing ergonomicness of pipeline milking machines - methodological approach. Leola, A., **Reinvee, M.**, Luik, M., Lensment, R.

5th International Conference Biosystems Engineering, 08.-09.05.2014, Tartu, Estonia, Utilisation of tactile sensors in ergonomic assessment of hand–handle interface. **Reinvee, M.**, Jansen, K.

6th International Conference Biosystems Engineering, 07.-08.05.2015, Tartu, Estonia, Concept of a low-cost device for quantitative evaluation of MSD risk factors. **Reinvee, M.**, Vaas, P., Kukk, H.

6th International Conference on Applied Human Factors and Ergonomics, 26.-30.07.2015, Las Vegas, USA, Applicability of affordable sEMG in ergonomics practice. **Reinvee, M.**, Vaas, P., Ereline, J., Pääsuke, M.

2nd International Conference Contemporary Ergonomics challenges in Europe, 26.11.2015, Riga, Latvia, Contemporary low-cost EMG, applications and quality. **Reinvee, M**.

7th International Conference Biosystems Engineering, 12-13.05.2016, Tartu, Estonia, Strength potential of lateral pinch activities: a work in progress. **Reinvee, M.**, Luik, V.

29th International Seminar of Ergonomics, 20-22.06.2016, Gniezno, Poland, Low-cost sEMG capture and feedback applications on Arduino platform. **Reinvee, M**.

7th International Conference on Applied Human Factors and Ergonomics; 27-31.07.2016, Orlando, USA, An electromyographic analysis of 'bend the tool not the wrist' concept on power tools. **Reinvee, M.**, Aia, S.

Human Factors and Ergonomics Society International Annual Meeting, 19–23.09.2016, Washington D.C., USA, Overview of Contemporary Low-cost sEMG Hardware for Applications in Human Factors and Ergonomics. **Reinvee, M.**, Pääsuke, M.

3rd International Conference Actualities of Ergonomics in Nowadays, 14.10.2016, Riga, Latvia, Ergonomics Development Tendencies in Estonia. **Reinvee, M**.

8th International Conference Biosystems Engineering, 11-13.05.2017, Tartu, Estonia, Ergonomic assessment of power tool's rotatable handle: a work in progress. **Reinvee, M**., Aia, S., Kukk, H., Jaaniste, J., Pääsuke, M.

International Symposium on Occupational Safety and Hygiene, 26.-27.03.2018, Guimarães, Porgutal, An investigation of forearm EMG normalization procedures in a field study. Aia, S., **Reinvee, M**.

31st International Seminar of Ergonomics, 23-25.05.2018, Gniezno, Poland, Feasibility of a low-cost microcontroller based lumbar flexion-relaxation phenomenon monitoring system. Gussev, V., Veraksitš, A., **Reinvee, M**.

VIIS VIIMAST KAITSMIST

PEETER PADRIK
FACTORS INFLUENCING THE QUALITY OF SEMEN FROM ESTONIAN HOLSTEIN AI BULLS, AND RELATIONSHIPS BETWEEN SEMEN QUALITY PARAMETERS AND *IN VIVO* FERTILITY
EESTI HOLSTEINI TÕUGU SUGUPULLIDE SPERMA KVALITEET, SEDA MÕJUTAVAD TEGURID NING SEOS *IN VIVO* VILJAKUSEGA
Professor **Ülle Jaakmaa**, Professor **Olev Saveli**
1. november 2019

TARMO NIINE
IMPACT OF GASTROINTESTINAL PROTOZOAN INFECTIONS ON THE ACUTE PHASE RESPONSE IN NEONATAL RUMINANTS
SEEDEKULGLAT TÕVESTAVATE ALGLOOMADE MÕJU MÄLETSEJALISTE ÄGEDA JÄRGU VASTUSELE NEONATAALPERIOODIL
Professor **Toomas Orro**, doktor **Brian Lassen**
21. november 2019

IVAR OJASTE
BREEDING AND MIGRATION ECOLOGY OF EURASIAN CRANE (*GRUS GRUS*)
SOOKURE (*GRUS GRUS* L.) PESITSUS- JA RÄNDEÖKOLOOGIA
Professor **Kalev Sepp**, vanemteadur **Aivar Leito**†, vanemteadur **Ülo Väli**
1. detsember 2019

BIRGIT AASMÄE
ANTIMICROBIAL RESISTANCE OF ESCHERICHIA COLI AND ENTEROCOCCI ISOLATED FROM SWINE, CATTLE AND DOGS AND MASTITIS PATHOGENS ISOLATED IN ESTONIA IN 2006–2015.
EESTIS AASTATEL 2006–2015 SIGADELT, VEISTELT JA KOERTELT ISOLEERITUD ESCHERICHIA COLI JA ENTEROCOCCUS'E PEREKONNA MIKROOBIDE NING LEHMADELT ISOLEERITUD MASTIIDIPATOGEENIDE ANTIBIOOTIKUMIRESISTENTSUS.
Dotsent **Piret Kalmus** ja professor **Toomas Orro**
13. detsember 2019

KAIA KASK
THE EFFECTS OF HEAT STRESS SEVERITY ON PHOTOSYNTHESIS AND VOLATILE ORGANIC COMPOUND EMISSIONS IN BLACK MUSTARD AND TOBACCO.
KUUMASTRESSI MÕJU MUSTA KAPSASROHU (*BRASSICA NIGRA* L.) JA VÄÄRISTUBAKA (*NICOTIANA TABACUM* L.) FOTOSÜNTEESILE JA LENDUVÜHENDITE EMISSIOONIDELE.
Professor **Ülo Niinemets**, vanemteadur **Astrid Kännaste**
5. märts 2020

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